Master's Degree in Industrial Technologies

Master's final project

Influence of active musculature on injury risk prediction on lateral crashes

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Madrid
July 2021

Declaro, bajo mi responsabilidad, que el Proyecto presentado con el título

Inlfuence of active musculature on injury risk prediction on lateral crashes

en la ETS de Ingeniería - ICAI de la Universidad Pontificia Comillas en el

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El Proyecto no es plagio de otro, ni total ni parcialmente y la información que ha sido

tomada de otros documentos está debidamente referenciada.

Fdo.: Arturo Pérez Calero Fecha: 13/ 07/ 2021

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INFLUENCIA DE LA MUSCULATURA ACTIVA EN LA PREDICCIÓN DEL RIESGO DE DAÑO EN COLISIONES LATERALES

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Entidad Colaboradora: Volvo Cars AB

RESUMEN DEL PROYECTO

1.- Introducción

Con la industria del automóvil encaminada hacia la conducción autónoma (AD), y con los *Advanced Driving Assistance Systems* (ADAS) ya en el mercado, es más probable que nuevas maniobras de evasión realizadas por estos sistemas se den en el tráfico rodado. Estas maniobras podrían aumentar el número de colisiones evitadas, pero también aumentarían las posibilidades de que otras colisiones no evitadas fueran precedidas por ellas.

Estas maniobras de evasión, como la frenada autónoma de emergencia (AEB) o el asistente de carril (LKA), podrían inducir movimiento en el ocupante justo antes de la colisión, cambiando las condiciones previas al impacto, lo que podría afectar a las lesiones resultantes de la colisión.

Por lo tanto, resulta interesante estudiar la influencia de la predicción del riesgo de lesiones cuando una colisión es precedida por una maniobra de evasión. Una de las herramientas utilizadas para analizar la cinemática del ocupante son los *Human Body Models* (HBM), que incluyen musculatura activa para replicar la respuesta de un ser humano ante los escenarios de carga con bajas aceleraciones que se producen durante las maniobras de evasión.

Por lo tanto, el objetivo de este Proyecto consistió en analizar la influencia de la predicción del riesgo de daño del ocupante al incluir la musculatura activa en un HBM expuesto a maniobras de evasión antes de la colisión.

Esto se llevó a cabo mediante la simulación de modelos con y sin musculatura activa, de modo que se pudiera realizar una comparación entre ambos.

2.- Metodología

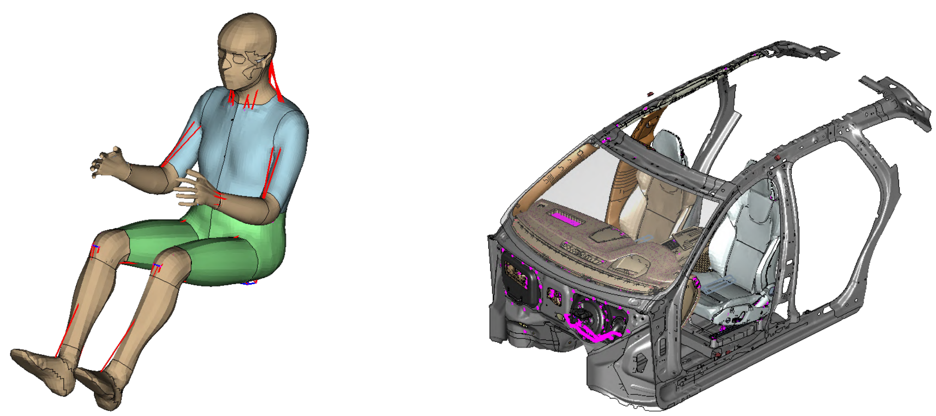
En este proyecto se realizaron simulaciones con modelos de elementos finitos para estudiar los diferentes escenarios de mabiobras de evasión y colisiones con un HBM con musculatura activa. El HBM fue situado en un modelo de un *sled*, proporcionado por *Volvo Cars*, en el asiento del pasajero. Se activó y desactivó la musculatura activa del HBM para comparar ambos casos, lo cual era el objetivo principal de este Proyecto.

2.1.- Configuración de las simulaciones

El desarrollo de este Proyecto requirió principalmente de 2 modelos: el HBM, que representó al pasajero del vehículo, y el *sled* que representó al propio vehículo sobre el que se realizaron las simulaciones.

En cuanto al HBM, se utilizó el *SAFER A-HBM v10.0*, que representa un varón de percentil 50. Este modelo fue desarrollado por *SAFER* y acaba de lanzar su versión 10.0. Por lo tanto, este proyecto también podría utilizarse para futuras verificaciones de esta nueva versión, ya que es una continuación del trabajo realizado en [1], donde se utilizó la versión 9 del *SAFER A-HBM*.

El sled consistía en la parte fronto-interior completa de un coche, que fue proporcionada por *Volvo Cars*. Ambos modelos se muestran en la *Figura 1*.



(a) SAFER A-HBM v10.0 en posición original.

(b) Modelo del *sled*.

Figura 1: Modelos utilizados en el Proyecto.

Primero se posicionó el *SAFER A-HBM* en el asiento. A continuación, se adaptó la espuma del asiento para adaptarla a la forma del HBM. Después se colocó el cinturón de seguridad y se incluyó todo el HBM con el cinturón de seguridad y la espuma del asiento actualizados dentro del modelo del *sled*. Por último, se definió la estrategia de activación de los sistemas de retención para cada caso de carga y se incluyeron los pulsos correspondientes a cada escenario de maniobra evasiva y colisión.

Las simulaciones se realizaron en *LS DYNA*. Además, se utilizaron *ANSA (BETA CAE Systems)* y *Oasys Primer (Oasys Software)* como preprocesadores para la configuración de las simulaciones, y *META (BETA CAE Systems)* y *LSPrePost (Livermore Software Technology)* como postprocesadores.

El proceso completo para ejecutar una simulación se muestra en la *Figura 2*. En primer lugar, se simularon 300 ms sin incluir pulsos para inicializar los controladores musculares. Justo después, se añadió el pulso de la maniobra de evasión, seguido del de de la colisión. Una simulación completa duró, por tanto, entre 1100 ms y 1450 ms, que se correspondía con un tiempo de computación de una semana aproximadamente.

Figura 2: Secuencia completa de simulación para un modelo activo.

Por lo tanto, para ahorrar tiempo de cálculo, sólo se simuló la secuencia copmleta para los modelos con musculatura activa. Al final de las maniobras de evasión de estas simulaciones, se guardó el estado de todo el modelo para extraer varios parámetros del HBM que se utilizaron como *inputs* para reiniciar los modelos pasivos desde ese punto.

Estos parámetros se añadieron a las simulaciones pasivas siguiendo la estrategia mostrada en la *Tabla 1*, donde la posición y la velocidad se refieren a la posición y la velocidad de cada nodo del HBM, la tensión se refiere a todas las tensiones de distintas partes del HBM y la señal PID se refiere a la última señal obtenida de los controladores musculares PID, que se utilizó para retener la tensión en ese punto de la simulación activa.

Esto se hizo para estudiar qué influencia tenían estos parámetros a la hora de replicar, mediante una simulación pasiva, la simulación activa correspondiente con el mismo caso de carga.

	Retained parameter			
Simulation	Position	Velocity	Stress	PID signal
0				
1	X			
2	X	X		
3	X	X	X	
4	X			X
5	X	X		X
6	X	X	X	X

Tabla 1: Matriz de estrategia para las simulaciones pasivas.

2.2.- Selección de maniobras

Como este proyecto se centró en colisiones laterales, sólo se simularon las maniobras evasivas que se esperaba que acabaran en una colisión lateral. Tras consultar [2], [3] y [4], se encontró que las maniobras más comunes antes de una colisión genérica podían simplificarse en cinco casos: frenar, cambiar de carril a la izquierda, cambiar de carril a la derecha, cambiar de carril a la izquierda mientras se frena y cambiar de carril a la derecha mientras se frena. Las combinaciones seleccionadas de maniobras evasivas y colisiones para este proyecto se muestran en la *Tabla 2*, done LT significa giro a la izquierda y RT, giro a la derecha.

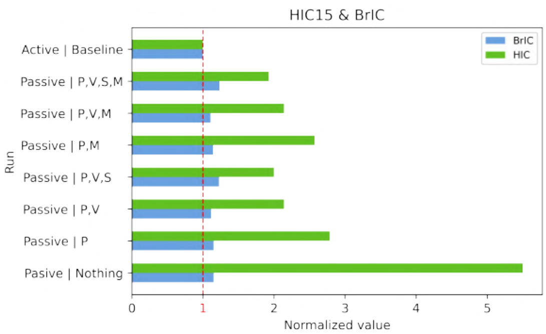
		Pre-crash maneuvers				
		B	LT	RT	LT + B	RT + B
In-crash	Far side	X	X		X	
	Near side	X				

Tabla 2: Combinaciones de maniobras evasivas y colisiones seleccionadas para el Proyecto.

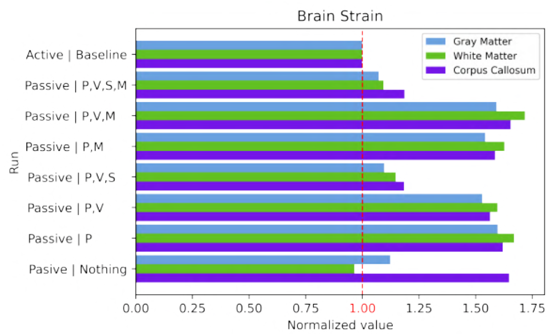
3.- Resultados

Las simulaciones pasivas mostradas en la *Tabla 1* se compararon con los modelos activos, que se trataron como simulaciones de referencia. Estas comparaciones se realizaron con el fin de estudiar la influencia de cada parámetro a la hora de replicar el comportamiento de la simulación de referencia correspondiente. Para ello, se extrajeron varios criterios de riesgo de lesión y modelos de deformación de los HBM de referencia y pasivos. También se llevaron a cabo análisis de correlación basados en las aceleraciones de la cabeza, el tórax y la pelvis. Aunque, por una cuestión de tiempo, no se simuló un modelo pasivo completo, las simulaciones pasivas con las correlaciones más altas se compararon con las simulaciones de referencia para estudiar una posible influencia de la musculatura activa en el resultado del riesgo de lesión predicho.

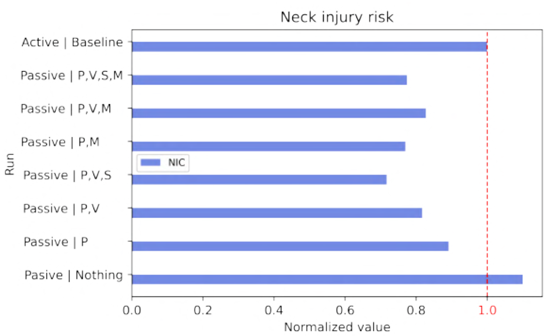
En las *Figuras, 3, 4, 5* y *6*, se muestran los resultados extraídos de uno de los grupos de simulaciones activas y pasivas. Los resultados se normalizaron con respecto a la simulación de referencia, ya que los valores reales son confidenciales.



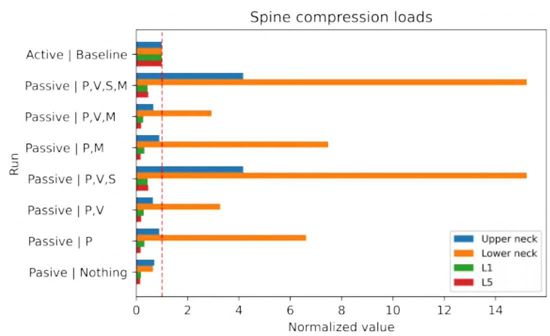
(a) HIC and BrIC. Normalizados.



(b) Brain strain. Normalizados.

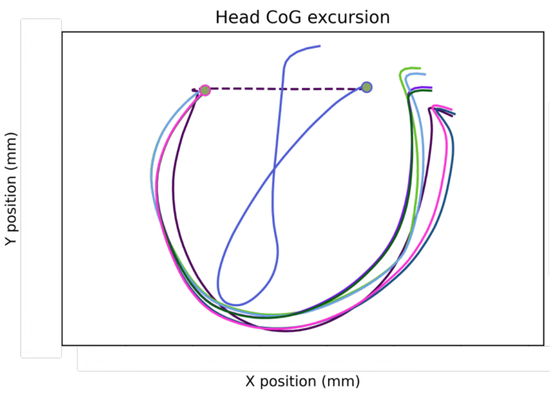


(c) Neck injury risk. Normalizados.

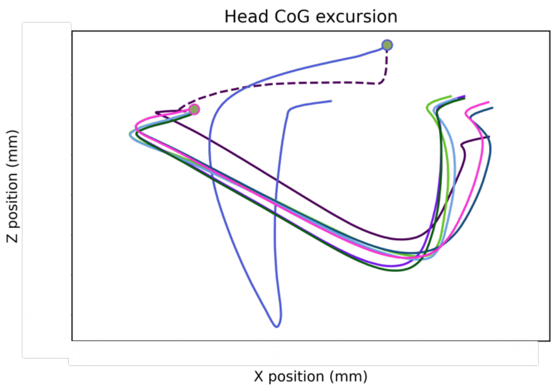


(d) Spine compression loads. Normalizados.

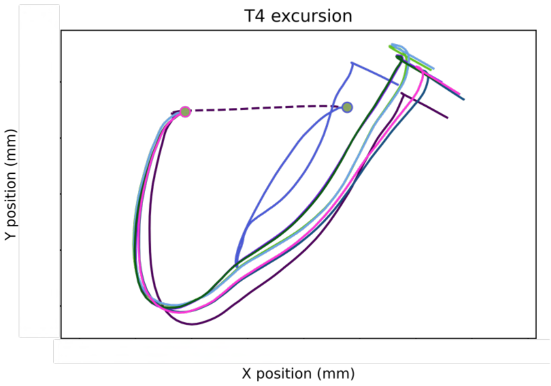
Figura 3: Criterios de lesión normalizados. Maniobra: frenado, colisión: lado lejano.



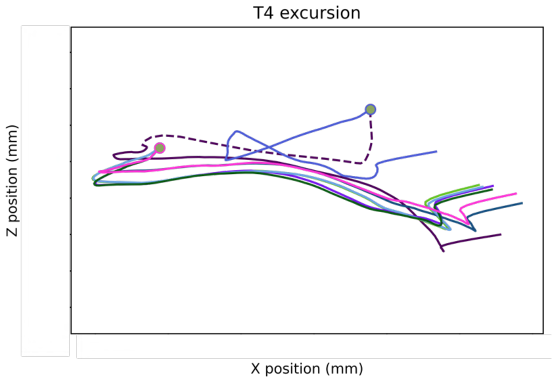
(a) Excursiones de la cabeza. Plano XY



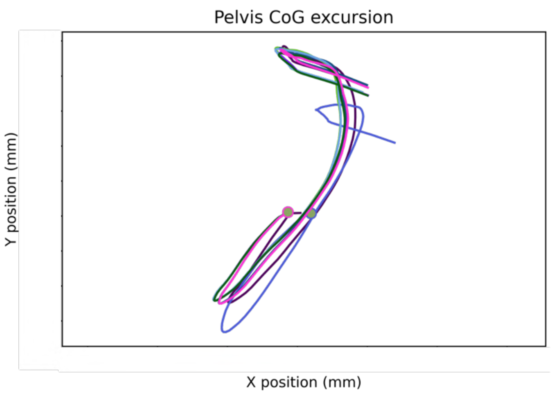
(b) Excursiones de la cabeza. Plano XZ



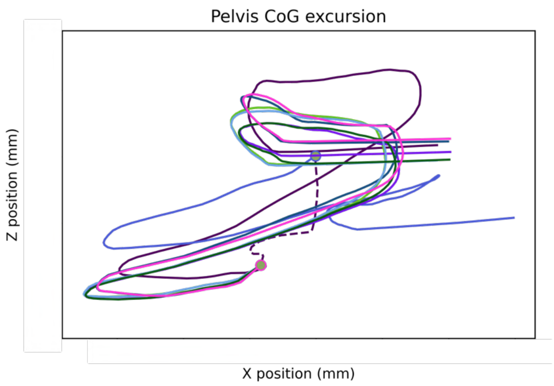
(c) Excursiones de T4. Plano XY



(d) Excursiones T4. Plano XZ



(e) Excursiones de la pelvis. Plano XY



(f) Excursiones de la pelvis. Plano XZ

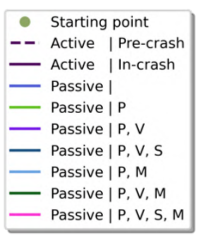
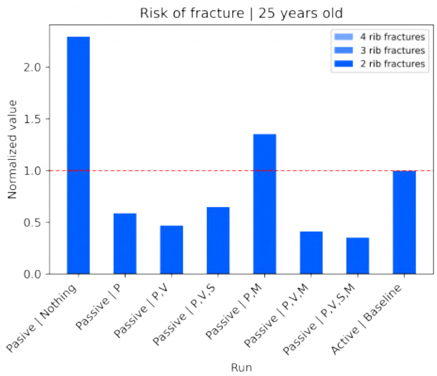
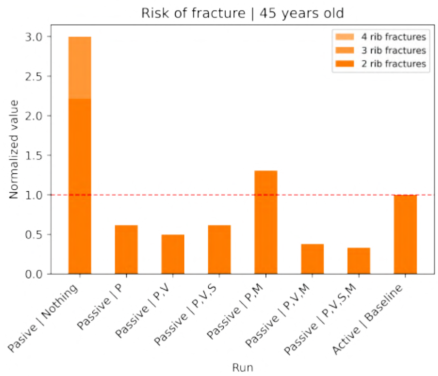


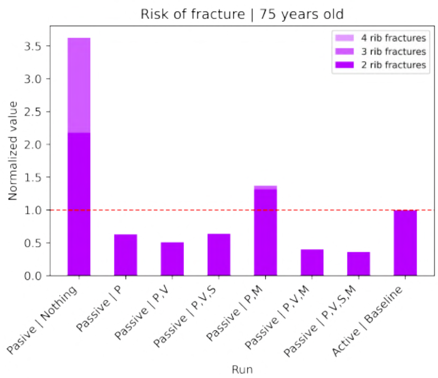
Figura 4: Excursiones de cabeza, T4 y pelvis. Maniobra: frenado, colisión: lado lejano.



(a) Riesgo de rotura de costillas normalizado. 25 años.



(b) Riesgo de rotura de costillas normalizado. 45 años.



(c) Riesgo de rotura de costillas normalizado. 75 años.

Figura 5: Riesgo de rotura de costillas normalizado. Maniobra: frenado, colisión: lado lejano.

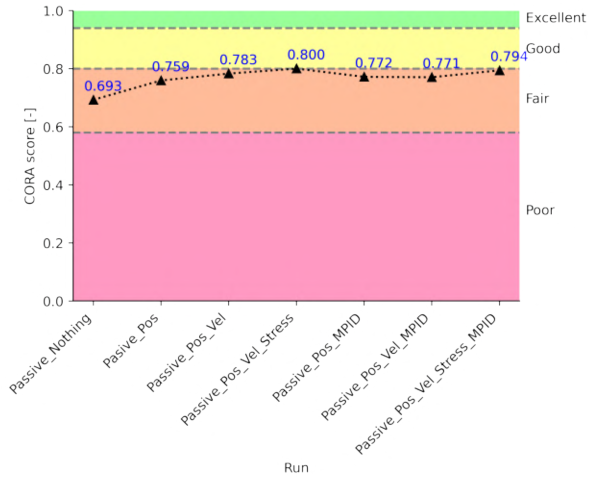


Figura 6: Resultados de CORA. Maniobra: frenado, colisión: lado lejano.

4.- Conclusiones

Los resultados concluyeron que, en general, cuantos más parámetros retenidos se añadían de una colisión pasiva a las simulaciones pasivas, más se acercaban éstas a las activas. El nivel de actividad muscular retenido antes de la colisión resultó mantener, y a veces disminuir, los resultados de la correlación con las simulaciones activas. Además, tuvo una influencia muy baja en las predicciones de riesgo de lesión. Por lo tanto, los resultados de este proyecto mostraron que este parámetro no sería de especial utilidad a la hora de replicar un modelo activo.

Esto sugirió que los músculos no tienen influencia durante la colisión[1], ya que el nivel de actividad muscular retenido tuvo una influencia baja o nula cuando se incluyó en las simulaciones pasivas. Esto era de esperar, ya que la fuerza desarrollada por los músculos podría ser considereada como despreciable en comparación con las fuerzas implicadas en caso de colisión [5].

Sin embargo, se comprobó que la posición era el parámetro retenido que tenía mayor influencia. Además de este parámetro, la velocidad y las tensiones también aumentaron la correlación. Por lo tanto, en base a los resultados de este proyecto, si un modelo activo quiere ser replicado por un modelo pasivo, la postura inicial es el parámetro clave a incluir, siendo la velocidad y las tensiones retenidas buenas adiciones para mejorar los resultados de encontrarse disponibles.

Por lo tanto, como comentario final, si se quiere replicar un modelo activo a partir de un modelo pasivo, el parámetro clave a incluir en el modelo debería ser la postura del HBM. Las velocidades y las tensiones también mostraron resultados mejorados al añadirlas a las simulaciones. El nivel de actividad muscular retenido no mejoró la correlación, lo que demuestra que la actividad muscular no influye en la predicción de lesiones durante un caso de carga en colisión.

[1] al contrario de lo que se concluyó en [1], donde los músculos resultaron tener influencia durante las maniobras previas a la colisión, es decir, casos de carga de baja aceleración.

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INFLUENCE OF ACTIVE MUSCULATURE ON INJURY RISK PREDICTION ON LATERAL CRASHES

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Director: López Valdés, Francisco José
Collaborating Entity: Volvo Cars AB

PROJECT SUMMARY

1.- Introduction

With the automotive industry heading towards Autonomous Driving (AD), and with Advanced Driver Assistance Systems (ADAS) already in the market, new avoidance maneuvers performed by these systems are more likely to appear in road traffic. These maneuvers could increase the number of avoided crashes, but also possibilities of having crashes preceded by them would increase.

These avoidance maneuvers, such as Autonomous Emergency Braking (AEB) or Lane Keeping Aid (LKA), could induce movement in the occupant right before the crash, changing the conditions of the impact, which may affect the injury outcome.

Therefore, it is interesting to study the influence of the injury risk prediction when having an in-crash preceded by a pre-crash avoidance maneuver. One of the tools used to analyze the kinematics of the occupant are Human Body Models (HBM), which lately include active musculature in order to replicate the response of a human being to the low accelerations load-cases that are caused during avoidance maneuvers.

Therefore, the objective of this Project consisted on analyzing the influence of the injury risk prediction when including active musculature on a HBM exposed to pre-crash avoidance maneuvers followed by in-crash scenarios.

This was done by comparing models that ran with and without active musculature.

2.- Methodology

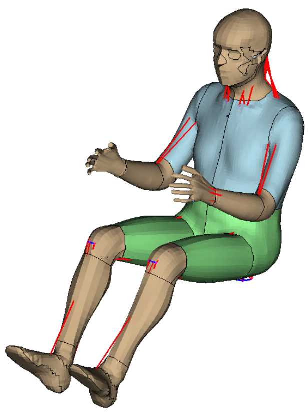
In this project, Finite Element simulations were carried out in order to study the different pre-crash and in-crash scenarios with a Human Body Model (HBM). The HBM was positioned in a sled model, provided by *Volvo Cars*, in the passenger seat. Active musculature of the HBM was activated and deactivated in order to compare both cases, which was the principal objective of this project.

2.1.- Simulations set-up

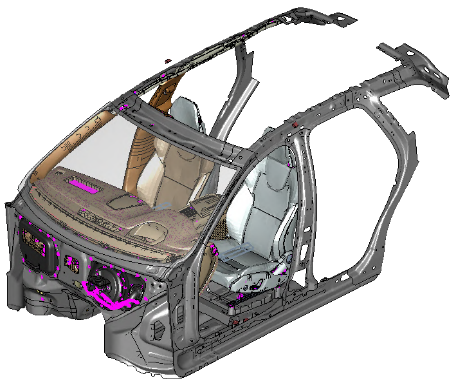
The development of this project involved two principal models: the HBM and the sled.

Regarding the HBM, *SAFER A-HBM v10.0* was used, which represents a 50^{th} percentile male. This model was developed by *SAFER* and has just released its 10^{th} version. So this project could also be used for future verifications regarding this new version, as this is a continuation of the work performed in [1], where version 9 of the *SAFER A-HBM* was used.

The sled consisted on the complete frontinterior part of a car, which was provided by *Volvo Cars*. Both models are shown in *Figure 1*.



(a) SAFER A-HBM v10.0 in original position.



(b) Sled model used for the Project

Figure 1: Models used in the Project.

The *SAFER A-HBM* was first positioned on the seat. Then, the foam of the seat was squashed in order to adapt to the shape of the HBM. Afterwards, the belt was routed and the whole HBM with the updated seatbelt and seat foam were included inside the sled model. Lastly, restraint systems triggering was adapted for each load-case, and pulses corresponding to each pre-crash and in-crash scenario were included.

Simulations ran in *LS DYNA*, which is a finite solver explicit solver software. Moreover, *ANSA (BETA CAE Systems)* and *Oasys Primer (Oasys Software)* were used as pre-processors for the set-up, and *META (BETA CAE Systems)* and *LSPrePost (Livermore Software Technology)* were used as post-processors.

The complete process for running a whole simulation is shown at *Figure 2.* Firstly, 300 ms with no pulses included were simulated to initialize the muscle controllers. Right after that, the pre-crash pulse was added, followed by the in-crash scenario. A complete simulation last, therefore between 1100 ms and 1450 ms, which took around one week to terminate.

Figure 2: Complete time process for an active simulation

Therefore, in order to save computational time, only models with active musculature were simulated with the complete sequence. At the end of these simulations' pre-crash maneuvers, the state of the whole model was saved in order to extract several parameters from the HBM that were used as inputs to restart the passive models from that point.

These parameters were added to the passive simulations following the strategy showed in *Table 1*, where position and velocity refer to each HBM's node position and velocity, stress refers to every solid, shell and beam stress in the HBM and PID signal refers to the last signal obtained from the PID muscle controllers, which was used to retain the stiffness at that point from the active simulation of the controllers.

This was done to study what influence did these parameters have in order to replicate the corresponding active simulation with the same load-case.

	Retained parameter			
Simulation	Position	Velocity	Stress	PID signal
0				
1	X			
2	X	X		
3	X	X	X	
4	X			X
5	X	X		X
6	X	X	X	X

Table 1: Passive simulations matrix strategy.

2.2.- Maneuvers selection

As this Project was focused on lateral crashes, only pre-crashes that were expected to end up with a lateral in-crash were simulated. By reviewing [2], [3] and [4], it was found that the most common pre-crash maneuvers could be simplified to five cases: braking, left lane change, right lane change, left lane change while braking and right lane change wile braking. The selected combinations of pre-crash and in-crash for this Project are shown in *Table 2*, where LT stands for left turn, and RT for right turn.

		Pre-crash maneuvers				
		B	LT	RT	LT + B	RT + B
In-crash	Far side	X	X		X	
	Near side	X				

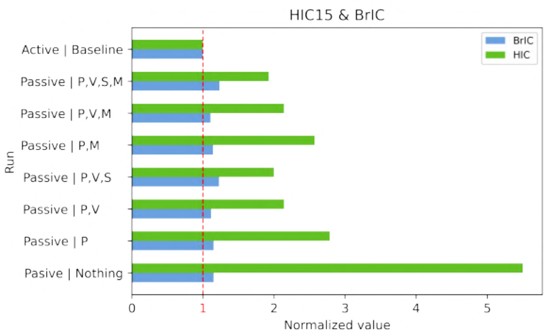
Table 2: Selected pre-crash and in-crash combinations

3.- Results

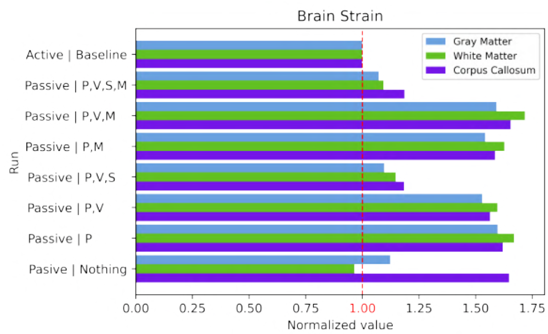
Passive simulations shown in *Table 1* were compared to the active models, which were treated as a baseline. These comparisons were done in order to study the influence of each parameter in replicating the behaviour of the corresponding baseline simulation. To do so, several injury risk criteria and strain models were extracted from the baseline and passive HBMs. Correlation analysis based on accelerations of the head, thorax and pelvis were also performed.

Although a complete passive model with a previous pre-crash simulation was not run due to a matter of time, passive simulations with the highest correlations were compared with the baseline to study a possible influence of the active musculature on the injury risk outcome predicted by both the passive and active models.

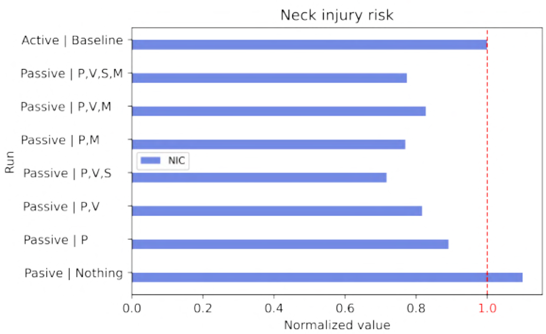
In *Figures, 3, 4, 5* and *6*, extracted results from one of the groups of active and passive simulations are shown. Results were normalized to the baseline, as the actual values are confidential.



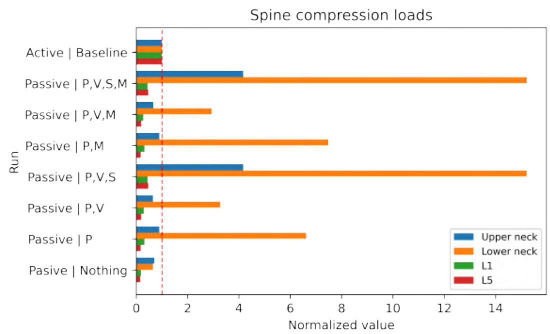
(a) HIC and BrIC. Normalized.



(b) Brain strain. Normalized.

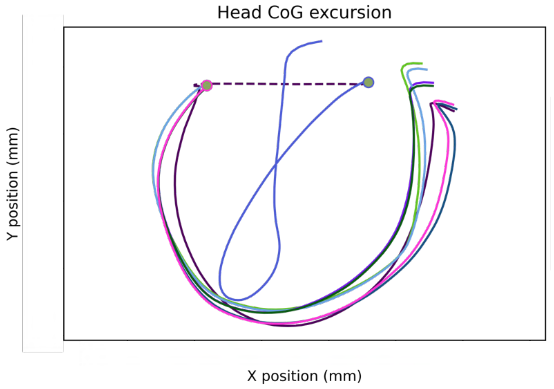


(c) Neck injury risk. Normalized.

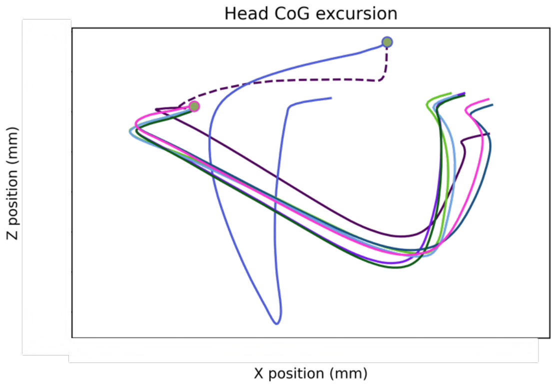


(d) Spine compression loads. Normalized.

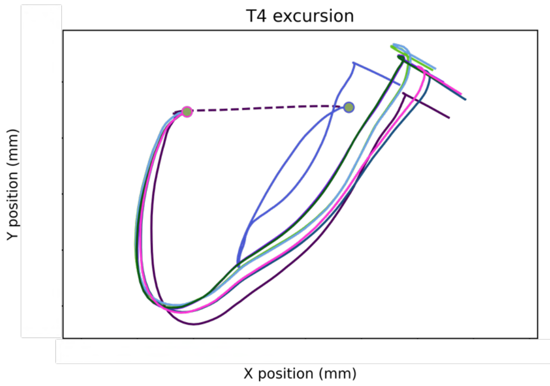
Figure 3: Normalized injury criteria. Pre-crash: braking, in-crash: far side



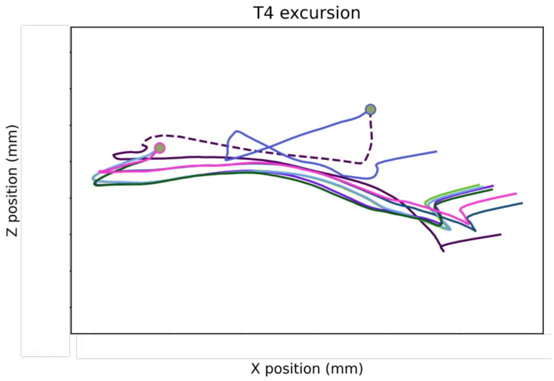
(a) Head CoG excursions *XY-plane*



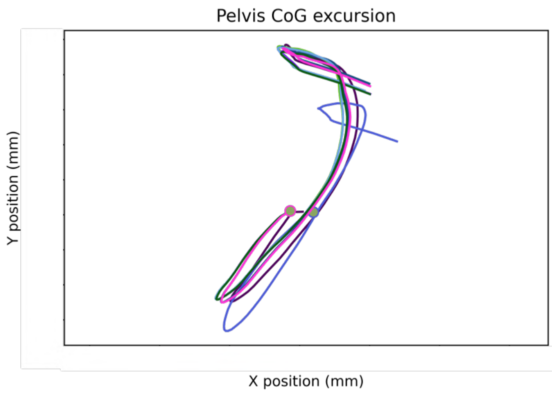
(b) Head CoG excursions *XZ-plane*



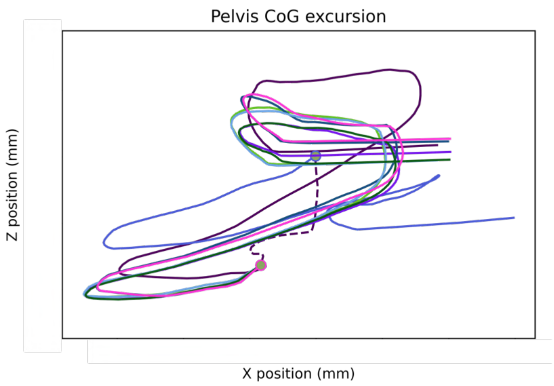
(c) T4 excursions *XY-plane*



(d) T4 excursions *XZ-plane*



(e) Pelvis CoG excursions *XY-plane*



(f) Pelvis CoG excursions *XZ-plane*

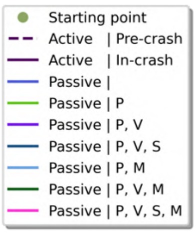
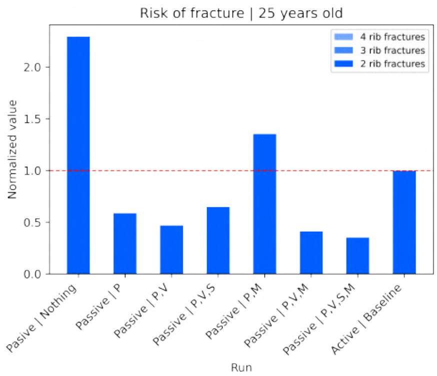
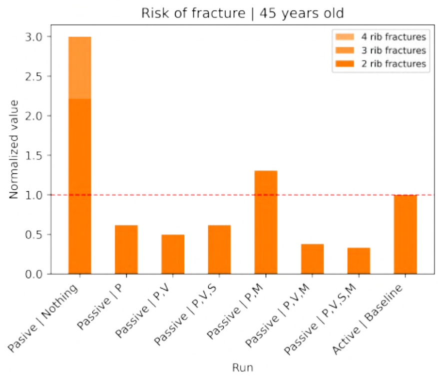


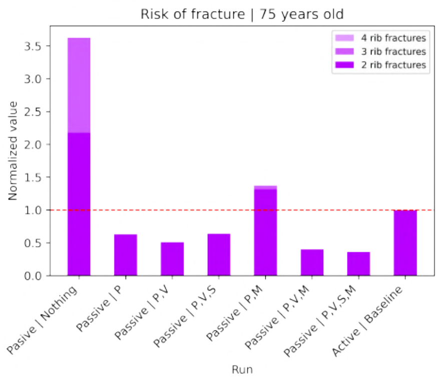
Figure 4: Head CoG, T4 and pelvis CoG excursions. Pre-crash: braking, in-crash: far side



(a) Risk of rib fracture. 25 years old. Normalized.



(b) Risk of rib fracture. 45 years old. Normalized.



(c) Risk of rib fracture. 75 years old. Normalized.

Figure 5: Normalized rib risk fracture. Pre-crash: braking, in-crash: far side

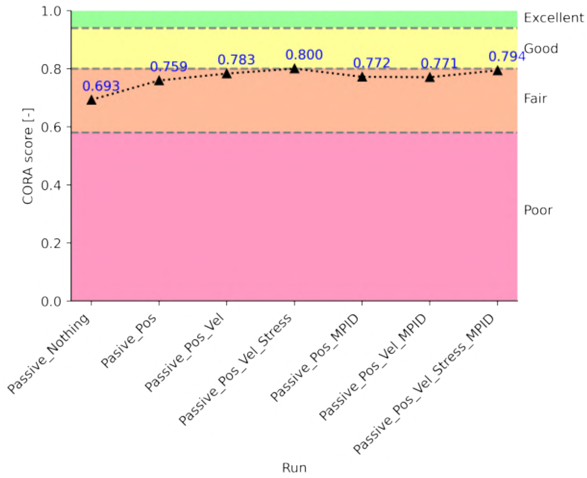


Figure 6: CORA score. Pre-crash: braking, in-crash: far side.

4.- Conclusions

Results concluded that, generally speaking, the more retained parameters added from a passive in-crash to passive simulations, the closer the passive simulations were to the active ones. Retained muscle activity level from pre-crash turned out to maintain, and sometimes lower, the correlation results with active simulations. Besides, it had very low influence regarding injury risk predictions. Thus, results in this Project showed that this retained parameter may not be a useful one to be used in order to replicate an active model.

This suggested that muscles do not have influence during the in-crash[1], as the retained muscle activity level had low or zero influence when included into the passive simulations. This was expected, as the force developed by the muscles could be considered as negligible in comparison to the forces involved in the event of an in-crash [5].

However, position was found out to be the retained parameter that had the greatest influence. In addition to this parameter, velocity and stresses also increased the correlation. Therefore, based on the results of this Project, if an active model wants to be replicated by a passive model, initial posture is the key parameter to include, with retained velocity and stresses being good additions for improving results if available.

Therefore, as a final comment, if an active model should be replicated from a passive model, the key parameter included in the model should be posture of the HBM. Velocities and stresses also showed improved results when adding them to the simulations. Retained muscle activity level did not improve correlation, proving that muscle activity does not influence the injury prediction during an in-crash load-case.

[1] On the other hand, in [1] it was concluded that muscles turned out to have influence during the pre-crash maneuvers, i.e., low acceleration load-cases.

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Influence of active musculature on injury risk prediction on lateral crashes

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Gothenburg, July 2021
Arturo Pérez

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Chapter 1

Introduction

The concern about human health has always been one of the most important issues humanity has faced. Historically, there have been several epidemics and pandemics that generated an enormous amount of deaths. These disasters are always treated as an urgent issue, due to the high level of mortality they cause. However, there is a big cause of death, which has been affecting humanity since approximately 125 years, which is traffic accidents.

Traffic accidents represented the 7^{th} cause of death (as it can be seen at the bottom right of *Figure 1.1* as *Road Inj*) in the world in 2019 [1]. This is a matter of concern, as road traffic has existed since the 19^{th} century.

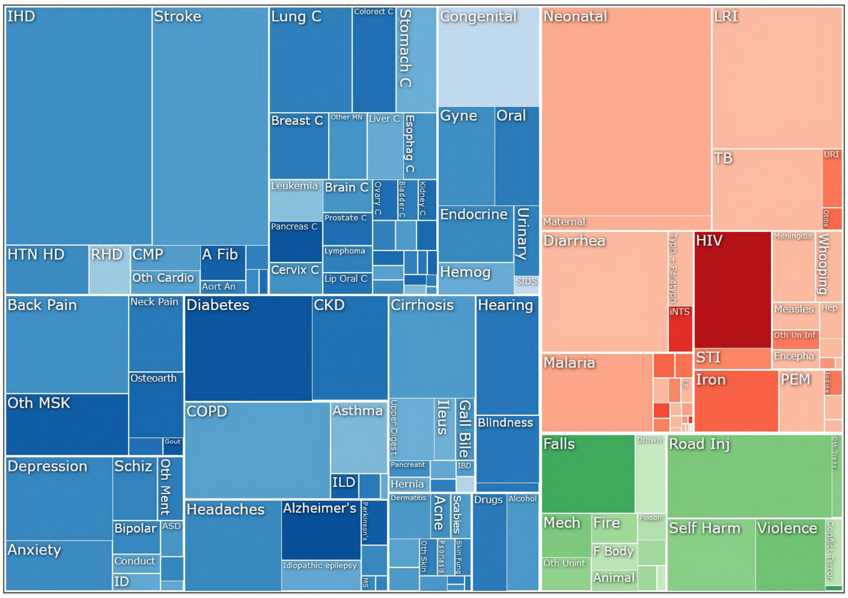


Figure 1.1: Principal causes of death in the world in 2019

Bridget Driscoll was the first road traffic mortal victim [2]. Since her death, a massive amount of people has lost their life in road traffic accidents, up to the current estimation that states that every year, there are 1.35 million of mortal victims due to traffic road injuries.

For instance, only in Spain, the number of fatal accidents increased until the year 1989 as it can be seen in *Figure 1.2*, reaching a record of 9344 people killed in road traffic accidents [3]. After that year, a big concern about road traffic deaths appeared and several measures were applied to reduce the number of victims.

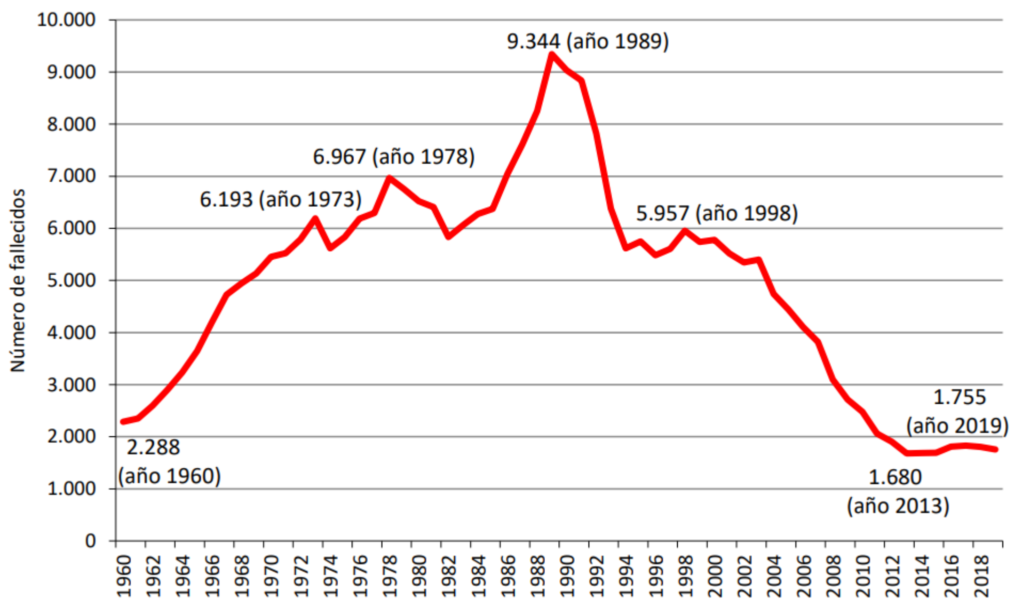


Figure 1.2: Evolution of fatalities in traffic accidents. Spain, 1960-2019 [3]

In 2019 the number of fatalities represents a 1% of the total number of road traffic accidents, injured hospitalized people represent a 6%, and injured non-hospitalized people represent a 93% [3]. This evolution shows the improvement in safety measures and it shows the current great concern about traffic injuries, which has brought up some global initiatives like *SUM4all* (Sustainable Mobility For All). They have the objective of reducing the number of traffic victims around the globe and they provide personalized recommendations in road traffic matter to make mobility safe, sustainable and accessible for everyone.

As mentioned before, not only the mortal victims present a concerning issue. Non-fatal victims are also a big problem, due to the hard recovery that some injuries require, as well as the money invested in those recoveries.

With all these facts, it seems reasonable to study in detail how the forces and accelerations generated in a crash affect the human body.

1.1 State of the art

Injury biomechanics is the science of forces acting on a biological structure and the mechanisms of resulting pathologies. Regarding automotive safety, there are three principal fields of study: in-crash safety, pre-crash safety, and integrated safety. In-crash safety is focused on mitigating injury in case of a collision, it acts once the crash has started. Pre-crash safety however focuses on avoiding the crash. Pre-crash safety systems actuate before the crash, operating over some controls of the car, helping the driver at the time of doing an emergency maneuver, or even avoiding the collision in a completely autonomous way. Lastly, integrated safety focuses on the integration between pre-crash and in-crash safety, studying the full sequence of a crash, starting from the pre-crash scenario, then the in-crash, and finally the post-crash.

The initial studies of biomechanics were focused on passive safety systems, developing the first seatbelt model back in 1940. The first official patent is attributed to the Swedish mechanical engineer from Volvo Cars, Nils Bohlin, who patented the three-point seatbelt that is used in cars nowadays [4]. The development of the concept of injury threshold started with John Stapp in the 1960s with his experiments in which he put himself on a sled and tested several decelerations over his own body to experiment how decelerations could affect the human body [5].

In 1975, Diemeter Adomeit and Alfred Heger wrote one of the most relevant articles for the development of restraint systems: *Motion Sequence Criteria and Design Proposalsfor Restraint Devices in Order to Avoid Unfavorable Biomechanic Conditions and Submarining* [6]. The position of the occupant and the way in which he is restraint are extremely important to avoid injury in the case of a collision. They studied different variables that affect the motion and kinematics of the occupant in a crash and developed some recommendations to avoid unfavorable biomechanics conditions and submarining (see *Figure 1.3*).

With this information, several 3-point belts were developed with different levels of technology to approach the kinematic proposed by Adomeit. Firstly, the belt implemented in cars was a standard 3-point belt that limited the frontal excursion of the occupant inside the car blocking the belt in case of a collision.

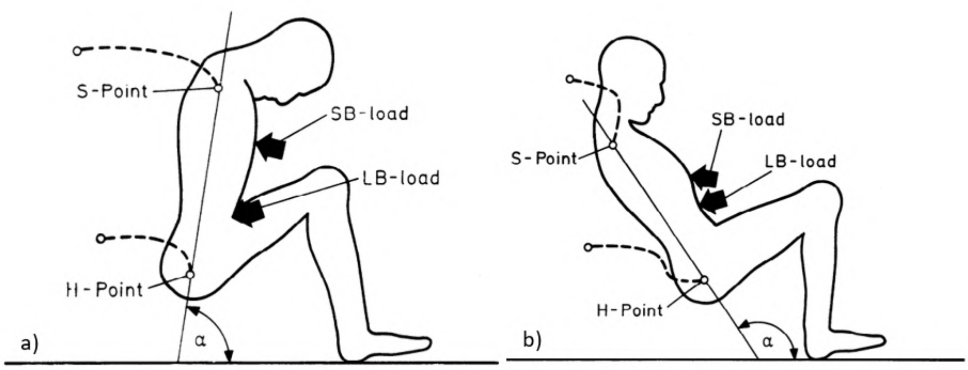


Figure 1.3: Trajectories and final position of 3-point belted passenger without classical ”submarining” (a) vs a “submarining” 3-point belted passenger (b) [6]

Modern seatbelts include pretensioners and load limiters. Pretensioners have the objective to adjust occupants into a safer seating position in case of a collision before the deployment of the airbag raising as well the compliance of the restraint system [7] using a pyrotechnic device that retracts both the lap belt and the shoulder belt, forcing the hip and the rib cage to move towards the back of the seat, in order to couple the occupant’s velocity with the vehicle’s velocity. Load limiters have the mission to control the maximum force the body is receiving from the seatbelt in the case of a crash. In the event of a crash, they protect the occupant by limiting the force transmitted to the occupant to a predefined maximum. This can be made with a torsion bar, which releases webbing gradually, so the force keeps a constant value below the injury limit.

Another passive safety system is the airbag. It was developed after the seatbelt and it consists of a bag positioned on the interior frontal part of the car, that inflates in the event of a crash. Its principal objective is to distribute the restraining forces across the occupant [8]. The airbag is therefore designed to actuate in combination with the seatbelt. Originally, airbags were only used for frontal crashes. However, there are currently many types of airbags situated in different parts of the car, in order to deploy them depending on the type of crash.

After the development of the airbag, the focus on automotive safety changed. Active safety systems or ADAS (Advanced Driver Assistant Systems) started to appear. They help the driver when maneuvering or braking. Some examples of these ADAS are the LKA (Lane Keeping Aid) or AEB (Autonomous Emergency Braking). As the last example, some of these ADAS may act in an autonomous way, helping the driver in case he/she cannot react in time. These kinds of systems can help to prevent several crashes developing some evasive maneuvers. However, even though their objective is to avoid a crash, they cannot ensure that the avoidance will be successful, and the car may end up colliding against another car or object. These maneuvers may affect the kinematics and posture of the occupant right before the crash, so it is important to study them in order to see if they influence the risk of injury to the occupant.

This is where the concept of integrated safety comes in. Every stage of a collision (pre-crash, in-crash, and post-crash) could be affected by the others, so it is essential to study their interactions. This is one of the main objectives of this study.

Historically, the study of collisions has been carried out by means of physical vehicle crash tests. To study how accelerations of a car collision might affect the passengers, anthropomorphic test devices (ATD) were developed (see *Figure 1.4*).

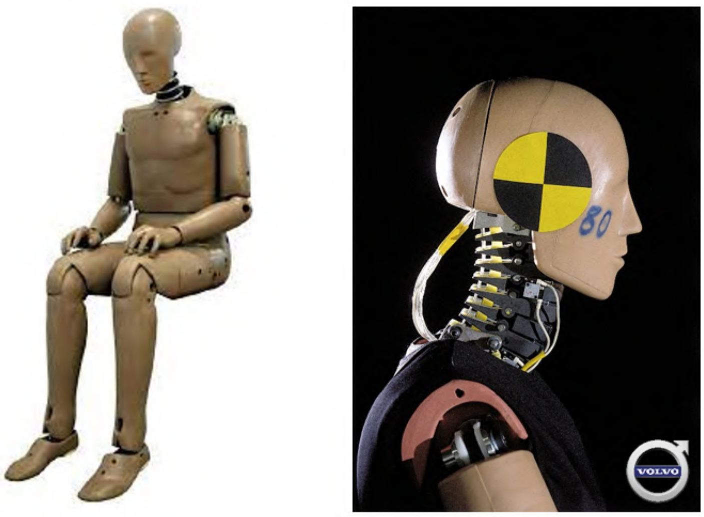


Figure 1.4: Harmonized Hybrid III 50th percentile Dummy

ATDs, also known as crash test dummies, are mechanical models created to surrogate the human body in replications of car crashes. They wear special instrumentation to monitor different parameters during the crash in order to predict possible injuries. Their principal problem is that they are not omnidirectional, ATDs are designed for a specific type of crash, and they do not have muscles, so they cannot recreate muscular activation.

ATDs are widely used for a great number of tests, validations, and studies. The most remarkable ones are the regulatory crash tests, which consist of standardized tests to evaluate the vehicle's safety quality for the occupant in the event of a car crash.

However, in order to study the occupant and driver kinematics using models capable of reproducing human muscular activation, human body models (HBM) were created [9] [10]. An HBM is a numerical model of a human body that is used for crash simulations. These simulations take place in different solvers, such as *LS DYNA*, *Madymo*, etc.

HBMs are being developed by different companies. One of the most famous one is the Total Human Model for Safety (THUMS) [11], developed by Toyota Motor Corporation (see *Figure 1.5*). HBMs may potentially be very important in injury biomechanics research.

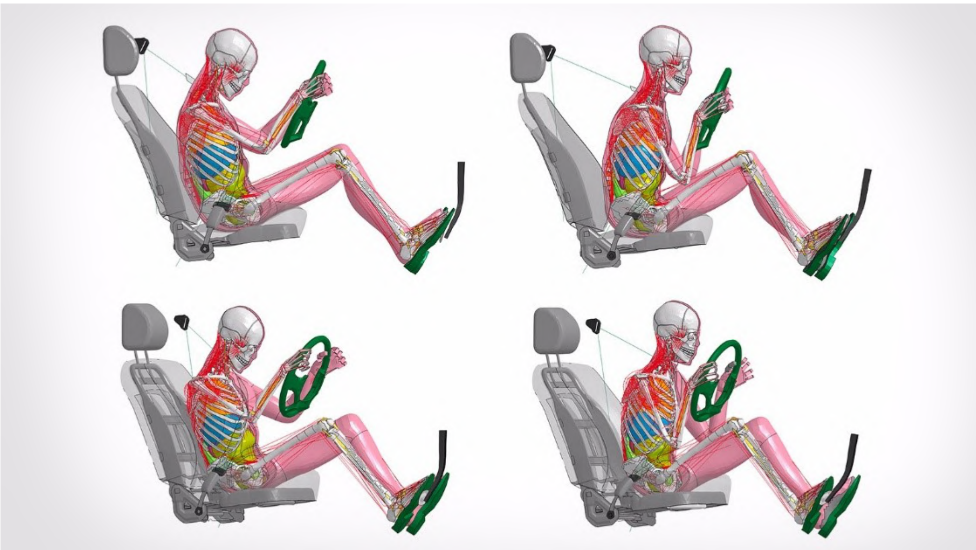


Figure 1.5: Toyota THUMS simulated in a frontal crash [11]

They are still in development, and several improvements have been made over the last few years. Some of these improvements include active musculature. Active musculature allows the model to react to certain accelerations and to try to maintain its original posture, generating an even more realistic scenario. One of the first active finite element HBMs was developed by SAFER [9]. Under the name of *SAFER-HBM*, this model was developed in 2009 and it has been evolving since then. Its first aim was to replicate a frontal impact simulation, which was validated with real-life volunteer data.

The most common aproach in how active musculature can be integrated in a Human Body Model, is by implementing 1 dimensional elements attached to the model. This elements are controlled by PID (proportional-integral-derivative) controllers, which can be based in angle deviations (APF) or length changes (MLF) [12] [13].

Under the name of SAFER-HBM, this model was developed in 2009 and it has been evolving since then (see *Figure 1.6*). Its first aim was to replicate a frontal impact simulation, which SAFER managed to simulate and validate with real life volunteer data.

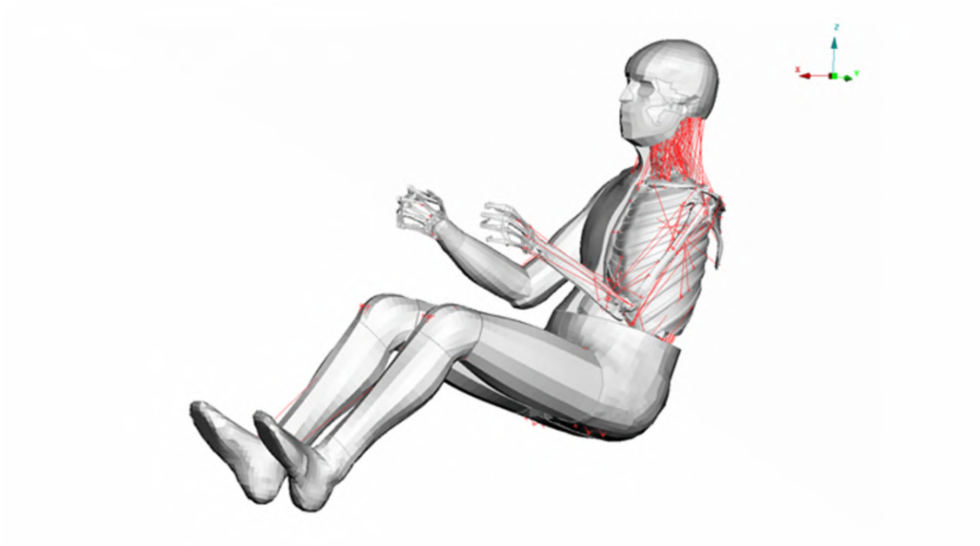


Figure 1.6: SAFER A-HBM [10]

Several studies state that occupants in a car are very sensitive to pre-crash maneuvers [13] [14]. As these maneuvers generate low-acceleration loads, it was found that musculature has a great influence over the kinematics of the volunteers. Therefore, it was found that A-HBM (Active Human Body Models) have a much bigger correlation with a real volunteer motion compared to passive models like ATD or standard HBMs.

1.2 Project aims

While moving towards Autonomous Driving (AD), and with ADAS still in the market, the possibilities of having a collision preceded by an avoidance (pre-crash) maneuver are currently increasing. This is the reason why studying the whole sequence of an avoidance maneuver followed by a crash has become very interesting.

This project was direclty based on a previous Master Thesis developed at the *Volvo Cars Safety Center* [15], in which the influence of different parameters captured at the end of the pre-crash in the occupant's response was studied.

Therefore, based on all the mentioned studies, the main objective of this project was to analyse the influence that the active musculature of the SAFER A-HBM has on the injury risk prediction during a crash. This was done by combining several pre-crash maneuvers with different in-crash scenarios, and running them with the active musculature turned on (active simulations) and off (passive simulations). This way a good comparison between passive and active can also be made in order to distinguish between the behaviour of the HBM during the pre-crash (low accelerations) and the in-crash (high accelerations).

Secondly, different parameters from the pre-crash were studied to verify which ones are useful to capture the full response of the HBM during the in-crash phase.

Chapter 2

Methodology

In this project, Finite Element simulations were carried out in order to study the different pre-crash and in-crash scenarios with the Human Body Model (HBM). The HBM was positioned in a sled model, provided by *Volvo Cars*, in the passenger seat. Active musculature of the HBM was turned on and off in order to compare both cases, which is the principal objective of this project.

These simulations require a considerable amount of computational power. In order to run them, *Volvo Car*'s cluster was used. Nevertheless, simulations took between 7 and 8 days to finish. That is the reason why a method to reduce time in some of them was developed, as it will be explained in *section 2.5.2*.

2.1 Software

Simulations ran using *LS DYNA R9.3.1*, which is a Finite Element explicit solver software. *ANSA (BETA CAE Systems)* and *Oasys Primer (Oasys Software)* were used as pre-processors for the set-up, and *META* (BETA CAE Systems) and *LSPre-Post (Livermore Software Technology)* as post-processors.

The analysis of the results, as well as the preparation of most of the data input to the models, was handled and/or created in *Python v3.8* through the *Spyder* environment.

2.1.1 LS DYNA brief overview

As the main tool used for this project was LS DYNA, a brief explanation of this powerful software will be made.

LS DYNA is a Finite Element explicit solver software, which means that it uses the Finite Element Method (FEM) for calculations. This method is a systematic calculation in which an infinite dimensions problem is discretized into finite elements defined by nodes. In these elements, the corresponding equations will be solved. In addition, by being an explicit solver, it calculates the state of the system at a time later than the current one. This is a great difference with the implicit solvers (such as the one ANSYS uses), which find a solution by solving the corresponding equations taking into account the current state of the system and the following one.

LS DYNA principally works with *KEYWORDS*. They have a pre-defined structure, and each of them is used for a different task. These tasks usually refer to very simple actions, such as creating a node, a shell (or any other element), defining contacts between two elements, setting the time interval in which the simulation will save information, etc.

The *KEYWORDS* are always written following the same order. First, the name of the *KEYWORD* is written beginning with an *, i.e. **NODE*. The following lines are called cards. Each *KEYWORD* has its own pre-defined cards, with a defined structure. Some of them are mandatory to fill, and others are optional. Besides, each *KEYWORD* has different options (with additional cards). For instance, they *KEYWORD* or defining a curve, **DEINFE_CURVE*, has the option to add a title to the curve, which would be **DEFINE_CURVE_TITLE*. In this case, an additional optional card to give a title can be used.

As a very simple example, the complete definition of a *KEYWORD* card for creating a node named 123456, at coordinates 1.00 (*X-axis*) 2.00 (*Y-axis*) 3.00 (*Z-axis*) will be the following.

```
*NODE
  123456              1.00              2.00              3.00
```

Figure 2.1: Example of a node definition

The resulting file after writing all the desired *KEYWORDS* is a text file with '.k' extension. Multiple files can be included inside other different files, which allows a better organization and understanding of the model, separating it into different modules that are easier to modify or update. All this files are included in a master file with '.key' extension, which is the one that is sent to the cluster in order to perform the simulation.

2.2 Models

The development of this project involved two principal models: the HBM and the sled.

Regarding the HBM, *SAFER A-HBM v10.0* was used, which represents a 50^{th} percentile male. This model was developed by *SAFER* and has just released its 10^{th} version, shown in *Figure 2.2*. So this project could also be used for future verifications regarding this new version, as this is a continuation of the work performed in [15], where version 9 of the *SAFER A-HBM* was used. Besides, *A-HBM* stands for *Active Human Body Model*, which means that it is an HBM that includes active musculature.

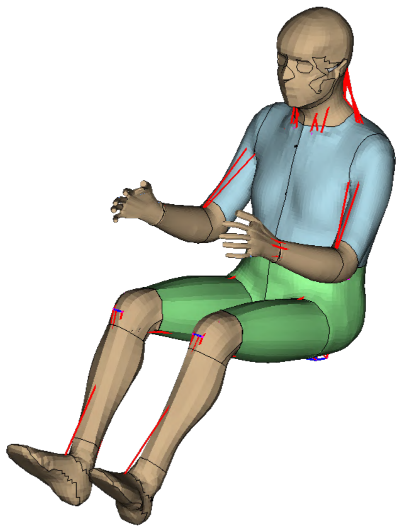


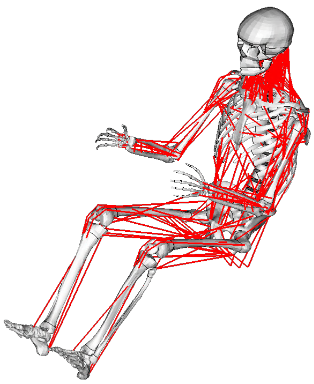
Figure 2.2: SAFER A-HBM v10.0 in original position.

The *SAFER A-HBM* has different closed-loop PID controllers for different body regions. It has also two different approaches, based on the angle deviation and the length change of the muscles [13]. The model consists on a principal file with all the nodes, beams, shells and more elements that form the human body model, another file that contains the muscles, and two more files that contain the PID controllers for the angles (APF) and the length (MLF). Finally, all these files are included in a master file, thus having the final model.

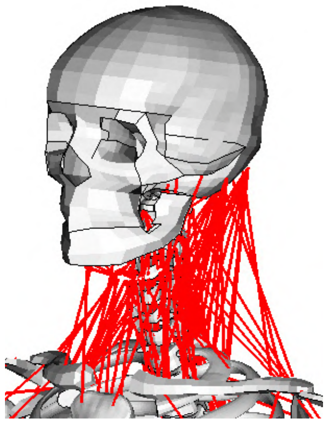
In this project, only the PID controllers related with the neck and the torso were used. The controllers for the arms were not activated, as the project was focused on the passenger. However, there is still future work to do when it comes to the driver occupant position, in which the arms controllers should also be included. In conclusion, the controllers activated for this project aimed to maintain the posture of the HBM (torso or lumbar active musculature) and to keep the head straight (neck active musculature).

In [13] it was found that a real passenger exposed to a lane change maneuver tends to keep the head aligned with gravity (vertical plane). In that study, it was also concluded that such behaviour is better represented by the APF controllers implemented in the *SAFER A-HBM*. Therefore, APF PIDs were the only approach in this project when it comes to controllers.

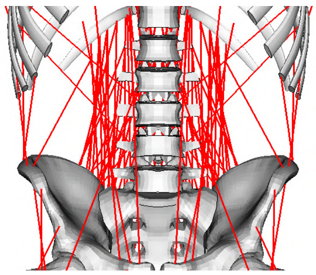
For better understanding, muscle elements are showed in red in *Figure 2.3*.



(a) All muscles



(b) Neck muscle elements



(c) Lower trunk muscle elements

Figure 2.3: Muscle elements visible in red.

Talking about the *SAFER A-HBM v10.0* itself, some minor numerical issues with ligaments were found. They will be mentioned in *Appendix* **??**. Moreover, data from the model was extracted after simulations to calculate some parameters, such as maximum brain strain, or ribs fracture risk.

Still work needs to be done with the *SAFER A-HBM v10.0* in order to validate it, as it is a very new model.

The sled consisted on the complete frontinterior part of a car (see *Figure 2.4*), which was provided by *Volvo Cars*. As the project is focused on the passenger, only the seat corresponding to that position was included in the simulations.

The sled was used as the environment were the pre-crash and in-crash pulses were applied in order to simulate the load cases of a real crash.

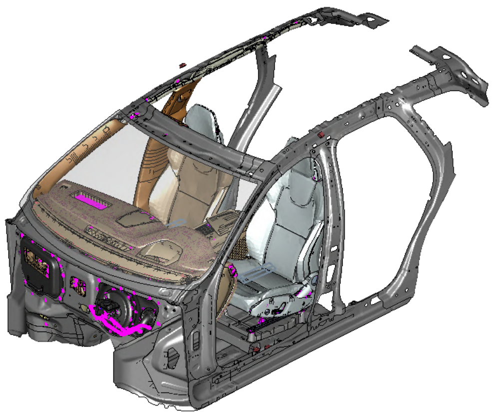


Figure 2.4: Sled model used for the Project

2.3 Maneuvers selection

When combining different pre-crash and in-crash maneuvers, there are almost infinite possibilities. Therefore, a previous research was done regarding which maneuvers are the most commonly executed before a crash [16] [17] [18]. This was done to reduce the number of simulations to a minimum, while still being representative of the real world.

After this research, it was found that the most common pre-crash maneuvers could be simplified to five cases: braking, left lane change, right lane change, left lane change while braking and right lane change while braking.

Regarding the in-crash scenarios, as this project is only focused on lateral crashes, two different types were selected to study: far side crash and near side crash [1].

Lateral crashes are principally related with intersections [18]. Frontal collisions, however, can occur in many other different types of situations [16] [17]. That is the reason why, after reviewing all the possible combinations, it was assumed that the pre-crash maneuvers more likely to happen before a lateral in-crash are: braking, left lane change and left lane change while braking. Finally, the selected combinations for those maneuvers are the ones marked in the *Table 2.1*, where LT stands for left turn, and RT for right turn. For this study, the braking maneuver represented an AEB and the side turns represented a LKS.

		Pre-crash maneuvers				
		B	LT	RT	LT + B	RT + B
In-crash	Far side	X	X		X	
	Near side	X				

Table 2.1: Selected pre-crash and in-crash combinations

Four different combinations were finally selected. They may seem few, but as commented above, lateral crashes are not that common (only in intersections [18]). If frontal crashes were analyzed in this project, there would have been much more possible combinations. Nevertheless, it was assumed that the selected combinations were enough to be representative for this study.

2.4 Simulations Set-Up

Set-up of the simulations was the task that took the longest to perform. It has several steps and processes, which will be briefly reviewed in this section.

2.4.1 HBM set-up

The initial step was to position the HBM in the passenger seat. In order to do so, the seat of the model was extracted and backrest and cushion angles were adjusted to a nominal position according to EURONCAP [19]. The result can be seen in *Figure 2.5*.

Followed by that, both the HBM and the seat were loaded in a *Primer* session

[1]This project is focusing on the passenger occupant, so when referring to far and near side crashes, they must be understood from the point of view of the occupant, i.e., they are flipped when comparing to the NCAP tests.

in order to adjust the posture of the HBM until getting a nominal [2] position for the passenger.

Figure 2.5: Prepared seat model

During this process, the seat was just used as a reference, so modifications were only made to the HBM. In fact, the HBM had penetrations with the foam of the seat to better represent the final posture once the foam had deformed due to the HBM's weight.

Once the the HBM's posture was defined, a simulation was run in order to position it. Cables were generated to pull each extremity and body part of the HBM into the new position. Therefore, one end of the cables is attached to the bones of the HBM, and the other is at the final position. This simulation took around 12 hours to run. Once it was done, the HBM was finally in the desired position to continue with the set-up. At this stage of the process, it is recommended to check the joints in the model, as they could have had relative displacement, resulting on its malfunction. That would require an additonal fix to correct the coordinates of the joins. However, in this case that was not neccesary, as the joints were perfectly updated from the simulation itself.

After this process, the HBM looked like it is shown in *Figure 2.6*. It can be appreciated that it has a different positioning than the original model showed at *Figure 2.2*. The most visible difference is the position of the arms, which must be on both sides of the thighs.

[2] This nominal poisiton refers to the standarized EuroNCAP protocol for dummy and HBM positioning. An example can be found at [19].

Once the HBM was completely positioned, it was time to squash the foam of the seat, due to as it was mentioned above, the HBM had penetrations with the it at this stage of the process.

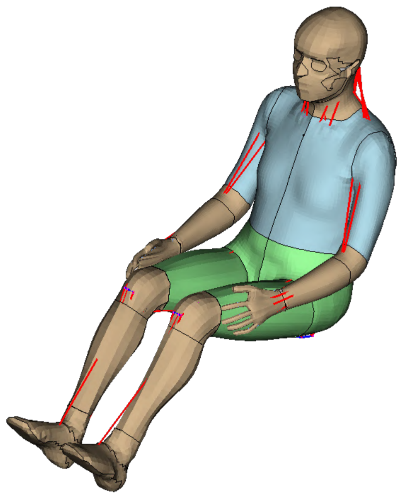


Figure 2.6: SAFER A-HBM v10.0 in nominal position

During this process, the HBM and the seat were loaded into *ANSA*, where a simulation was run to squash the foam of the seat in order to fit the position of the HBM. By doing this, there are no intersections between the HBM and the foam anymore, as the foam has already been squashed.

After having the HBM and the seat completely positioned and prepared, the next step was to route the belt. In order to do this step of the process, the HBM, the seat and the seatbelt model were loaded in *Primer*. This is a delicate process in which the routing of the baseline seatbelt has to be adapted to the position of the HBM. Contacts should be also defined between the belt webbing and the HBM. In this case, they were defined for the neck, chest, pelvis, thights and upper-arms, allowing penetrations with the lower-arms and hands for future stages of the process, as there could be problematic intersections between these parts.

Once the routing was done, the corresponding file was outputted from the model and included to the original seatbelt model replacing the original routing. At this stage, the HBM, seat and seatbelt are completely ready to be introduced into the sled model. *Figure 2.7* shows how the seated HBM with the routed seatbelt looks like.

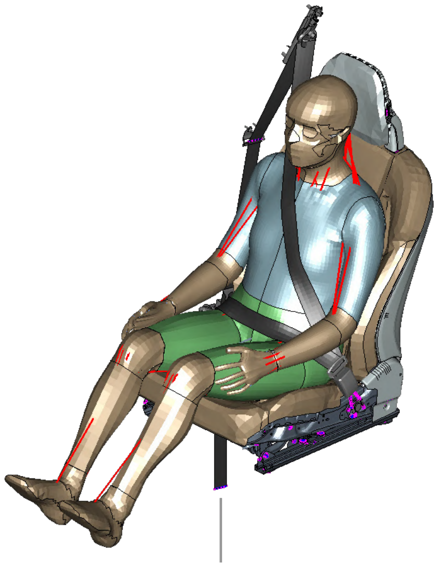


Figure 2.7: Seated HBM with seatbelt route updated

2.4.2 Sled set-up

By including the updated HBM, seat and seatbelt to the sled, the complete physical model is set-up. Still a lot of other parameters, such as materials and contacts definitions, including the different airbags and other parts of the sled, etc. had to be added. To do this, the sled model was included in a previous complete simulation file, which had the majority of those parameters already defined. It only was necessary to include the IC (interior courtain airbag) and SAB (side airbag), as they were used in for the near side crash. Regarding the contacts, they needed to be modified, as the ones from the previous simulation were defined for the *SAFER A-HBM v9*.

2.4.3 Other parameters & inputs

Once the complete model was prepared, there were still some inputs to define, which principally are the crash pulse to be applied and the strategy the restraint systems triggering.

**BOUNDARY_PRESCRIBED_MOTION_RIGID* was used for applying the pulses. This *KEYWORD* requires a previous definition of an speficic *PART* of the model where to apply the pulses, as well as the corresponding curves for each of the 6 degrees of freedom (3 translational and 3 rotational curves).

Concatenation of several curves can be made by the use of more than one of these cards, and by defining the "birth" and "death" of each one of them. This was precisely what was done, as pre-crash and in-crash pulses were obtained from separate curves.

The braking pre-crash maneuver was reused from [15]. Moreover, in order to replicate the same sequence of the mentioned project, an additional simulation was run for this maneuver, concatenating it with the same frontal crash. This was done as a check simulation, in order to verify that the results and motions were the same as the ones in the mentioned project with the *SAFER A-HBM v9*.

When it comes to the rest of the pulses used for this project, recorded lane change pulses from the test performed in [13] were utilized for the lateral pre-crash motions. This pulses consisted on a complete left lane change with and without braking. They both lasted for more than 3 seconds, which was way more than the aimed time for pre-crash maneuvers in this project. Thus, they were divided and only a part of them was finally used. The selected interval of time had to be short enough not to require too much computing time, but long enough to cause a lateral displacement that could justify this maneuver as evasive. Finally, an intermediate interval time of 850 ms was selected, causing a lateral displacement of around 1.5 m.

For this project it was not necessary to run with right lane change pulses. However, for other ongoing projects that needed them, these were generated by flipping the recorded left lane change pulses in *Y-axis* for translations and in *X* and *Z-axis* for rotations.

Regarding the in-crash pulses, they were also extracted from a pulse recording of a far side crash test (from the passenger point of view). Therefore, in order to apply the near side pulse, the original far side pulse was also flipped (same procedure as with the lane change) in *Y-axis* for translations and in *X* and *Z-axis* for rotations.

All this pulses were obtained from a global coordinate system, being the initial position of the car the same for all of them. This was a problem when trying to concatenate the pre-crash and in-crash pulses, as the global position of the car changed during the pre-crash maneuver, and it did not match with the initial position of the in-crash.

This could have been solved through different methods. However, the selected one consisted on first running a simulation applying only the in-crash pulse in global coordinates only to the BIW[3]. This way, curves in a local coordinate system created for the sled could be extracted.

Then, a new simulation was run applying the pre-crash in the global coordinate system and concatenating it with the extracted local in-crash pulse, applying it to the sled local coordinate system. Finally, both pulses from the pre-crash and in-crash were extracted as a longer global coordinate pulse from this simulation, thus having the concatenation completed.

It was important to keep in mind that for this method, the in-crash pulses were preceded by a pre-crash maneuver, which meant that the in-crash had initial velocities. The easiest fix for this was to apply the in-crash pulses as accelerations, so that they would add the in-crash motion to the given kinematics from the end of the pre-crash.

In this project, the final concatenated pulses applied to the definitive simulations were all accelerations (including both the pre-crash and in-crash, as well as the initialization time, which will be explained in the following section.

As a final comment, the previous method for pulse concatenation was not necessary for the pre-crash braking scenarios, as they only consisted on longitudinal deceleration, which did not modify the final pre-crash orientation of the sled. In addition, some of the resulting curves from intermediate pulse simulations needed to be filtered with a CFC60 filter, as the resulting curves were too noisy.

2.5 Simulations

After all the preparations mentioned in the previous section, the models were ready. To pursue the objective of this project, simulations with and without active musculature were set up. This means that for each pre-crash and in-crash combinations, there were two different types of simulations. Hereinafter, simulations with the active musculature turned on during the in-crash will be referred to as active simulations, whereas simulations with the active musculature turned off during the in-crash will be called passive simulations. Nevertheless, active musculature was only turned on and off during the in-crash, while the pre-crash always run as active.

These simulations required lots of computational time, and therefore a method to

[3]The reasons to run these simulations only with the BIW (Body In White) were on one hand, for them to run faster, and on the other hand, to apply the pulse only to a rigid body that is less sensitive to high peaks of accelerations.

try to reduce the computational time to a minimum needed to be developed.

After considering different options, the final method consisted on running the initialization and pre-crash maneuver only once, as they were common for the active and passive simulations. This way, the generic approach was to run the entire active simulation for a certain pre-crash and in-crash combination, and then restarting the simulation by the end of the pre-crash in order to run only the in-crash but this time as passive.

2.5.1 Active simulations

Active simulations were the ones that run the entire sequence of pre-crash and in-crash. As active musculature was used since the beginning of the simulation, the controllers of the active muscles needed at least 250 ms to initialize, for the model to reach equilibrium while being affected by gravity and finish adjusting to the seat. In order to do so, an initialization time of 300 ms was given to every active simulation.

This means that for the 300 first milliseconds, a zero acceleration pulse was applied to the model in order to maintain it static throughout the muscles initialization.

After that initialization period, the pre-crash pulse, instantly followed by the in-crash pulse, was applied. The pre-crash for the braking maneuver had a duration of 500 ms, while both lane changes had a duration of 850 ms. Lastly, the lateral in-crashes had a duration of 300 ms.

This leads to simulations with a duration of between 1100 ms and 1450 ms. The whole process can be seen in *Figure 2.8*.

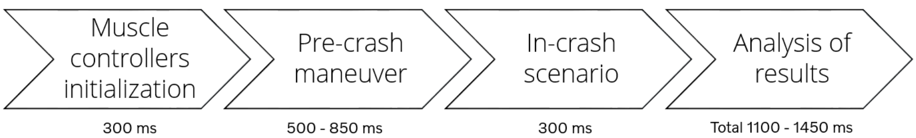


Figure 2.8: Complete time process for an active simulation

The major issue when running these simulations was to be able to restart them from a certain point in order to be able to re-run the in-crash as passive[4].

[4]As mentioned above, passive in-crash refers to an in-crash run with the HBM's active musculature turned off.

This could also have been solved through many methods. However, for the scope of this project, a good approach was to save all the information of the model at a certain time of the simulation using **CONTROL_STAGED_CONSTRUCTION*. This *KEYWORD* saves the information of all the model at a given stage of the simulation. Therefore, the simulation had to be divided into two stages using **DEFINE_CONSTRUCTION_STAGES*. The first stage was define from the beginning of the simultation to the end of the pre-crash maneuver, and the second one was defined from the beginning of the in-crash (or end of the pre-crash) to the end of the simulation. All these times were parameterized, so the only thing to do about them was to update the initialization, pre-crash and in-crash duration in the heading of the file.

The extracted information at the end of the pre-crash consisted on the updated position of every node in the model, as well as the retained velocity of each node and the retained stress of each solid, shell and beam. All this information is saved in a *Dynain* file, which is a text file that for this specific case contained millions of lines of code.

Regarding active simulations, this was the entire generic process necessary to run them. Each one of them took around 7 days to run using 120 cpus per simulation. While they were set up and running, the rest of the work regarding preparation of the passive simulations was developed.

2.5.2 Passive Simulations

Passive simulations were a restart of the active simulations from a previous point, which was the end of the pre-crash. Therefore, although only the in-crash was simulated, the complete sequence of a passive simulation was the exact same initialization and pre-crash maneuver run with the active musculature turned on (as shown in *Figure 2.8*), and followed by the in-crash scenario with the active musculature turned off.

The model was the same as the one used for the active simulations. However, changes needed to be done. To begin with, all the parameters related with time values had to be changed. As everything was parameterized, initialization and pre-crash times were set to zero, while the in-crash continued to be 300 ms. This way, the passive simulations only had a duration of 300 ms.

The next step was to include the parameters retained in the *Dynain* file from the end of the active simulations' pre-crash. This retained parameters were extracted separately, which allowed comparisons with the active models. The objective of this analysis was to detect which retained parameters affected the most to the output of the passive simulations in order to replicate as close as possible the results of the active simulations, which was the second principal aim of this project, as mentioned in *section 1.2*.

This way, the retained parameters to analyze were: HBM's posture, HBM's velocities, HBM's stresses and HBM's retained muscle activity level at the end of the pre-crash. In order to properly analyse their influence in results, a matrix simulation strategy was created, with different combinations of these parameters, shown in *Table 2.2*.

The posture of the HBM was extracted by updating the position of each node at the end of the pre-crash, which will have changed from the nominal position at the initialization phase. The velocity of the HBM was extracted as the velocity of each node at the specified time of the active simulations. The stress retainment referred to solids, shells and beams stress of the HBM. Lastly, The retained muscle activity level refers to the last PID signal of the controllers, which means that it will maintain the muscles with the same stiffness they had at the end of the pre-crash, but the simulation was still passive, due to there is no direct control of the muscles, i.e., the active musculature is still off.

	Retained parameter			
Simulation	Position	Velocity	Stress	PID signal
0				
1	X			
2	X	X		
3	X	X	X	
4	X			X
5	X	X		X
6	X	X	X	X

Table 2.2: Passive simulations matrix strategy.

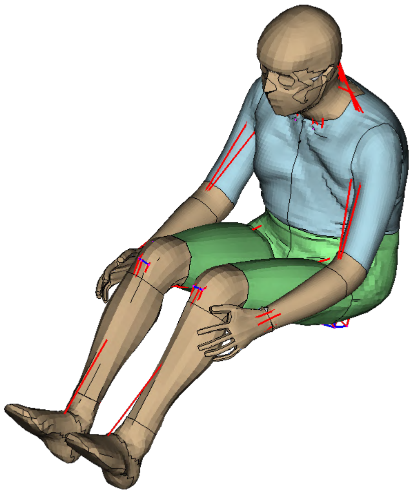
Firstly, a simulation called zero was run with no retained parameters, i.e., HBM in nominal position with no velocity or other parameters, just like the initial state of the active simulations. Then position, velocity and stress were added progressively through simulations 1 - 3. Simulations 4 - 6 were as the three previous ones, but with the last PID signal included. This strategy of combining different input parameters was based on results from [15], which concluded that the parameter with the most influence when it comes to frontal crashes preceded by a braking pre-crash maneuver was the position of the HBM. In this Project, different pre-crash and in-crash scenarios were studied. However, it was assumed that the mentioned strategy was a good starting point.

Thus, 7 different passive simulations were run for each in-crash, which gives a total of 28 total passive simulations and, counting with the 4 active simulations, a total of 32 simulations were run for this Project.

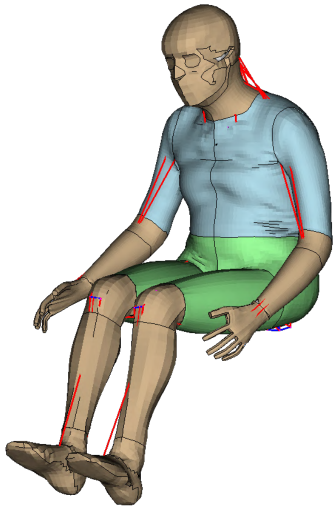
The first three parameters were extracted from the *Dynain* file generated from the corresponding active simulations. From this gigantic file, data had to be extracted and handled in order to make the extracted parameters to be compatible inputs for *LS DYNA*. For this purpose, *Python* was used. Data in the *Dynain* file was complex, with different formats within the same document and it gave some compatibility problems when it came to retained stress in some of the shells of the HBM.

The posture of the HBM was extracted as the position of each node and it was directly substituted in the HBM files when the simulation run. When including the posture of the HBM, it needed to be taken into account that the new posture also affected the seat and the seatbelt, which meant that their new position had to be updated too. Retained position for the *SAFER A-HBM* is shown in *Figure 2.9* .The velocity of the HBM was extracted as the velocity of each of its nodes, and it was included as an extra *LS DYNA* file with **INITIAL_VELOCITY_NODE*. When doing this, the initial velocity of the sled had also to be included to keep the appropriate relative velocity between car and occupant. This was included with **INITIAL_VELOCITY*, applied to every rigid body in the model. The stresses were extracted from solids, shells[5] and beams, and were included in the model in a separate *LS DYNA* file by the use of **INITIAL_STRESS_SOLID*, **INITIAL_STRESS_SHELL* and **INITIAL_STRESS_BEAM*. In addition, stresses in the foam of the seat were also included when the position of the HBM was included. This was done by calculating the new stresses in the foam of the seat comparing the geometry of the undeformed foam with the current one deformed by the new position of the HBM.
Regarding the last PID signal for the muscle controllers, it was not available from the *Dynain* file, so it had to be extracted directly from the output curves of the PID controllers located in the binout of each simulation. *META* was used for this task, by making a value report at the end of the pre-crash for every PID signal curve and

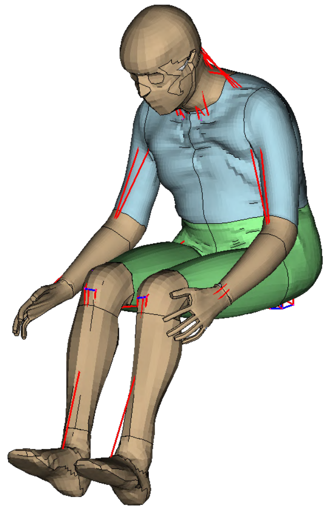
[5]The stresses of some of the shells could not be extracted in some minor parts of the HBM due to a format incompatibility between the *Dynain* file and *LS DYNA*. Dynamore was notified about this and is already working on solving this issue.



(a) Retained positon. Braking



(b) Retained position. Left lane change



(c) Retained position. Left lane change and braking

Figure 2.9: Retained positions for the *SAFER A-HBM*.

outputting it to a file that was afterwards handled in *Python*.

After completing the handling of all the data, it was time to prepare the rest of the model to run the simulations. It should be noted that the model had moved during the pre-crash, so when trying to restart the simulation to continue with the in-crash, the HBM, seat and seatbelt were not in the original position, due to when a new simulation starts, the sled and the rest of the elements in the model start from their defined initial position. This means that for simulations where posture and position were included, the HBM, seat and seatbelt had to be moved back into the sled. As the model also has rotations during the simulation,a transformation to these elements needed to be implemented. By using **DEFINE_TRANSFORMATION*, three nodes were created in the sled as a reference and three existing nodes of the sled were given as target nodes. Then when loading the simulation, the transformation measured where the target nodes were before and after compared to the reference ones, and then it calculates a 3-D transformation in order to bring back the target nodes to the reference nodes. By applying this transformation to the HBM, seat and seatbelt, these three elements were back in the original position and ready to run.

After this, the belt had to be re-routed in *Primer*, as it was not possible to re-initialize its state and running it from a certain point from the active simulations.

After all this process, some minor checks needed to be done in order to verify that all the connections between the rigid bodies were still well defined after the transformation of the HBM, seat and seatbelt.

After some checks, it was found that some ligaments in the feet and left knee were failing when trying to reinitialize them, due to they broke during the pre-crash in the active simulations. They had to be reunited by interpolation, or erased from the model in order to continue with the process. Also, due to a problem with contacts, two ligaments in the hip were initially penetrating the pelvis, so coordinates of those nodes needed also to be changed to solve those initial penetrations.

This process and fixes had to be done for each of the 28 passive simulations for this Project. Therefore, a *Python* script was developed in order to set up all the passive simulations from a given active simulation and its corresponding *Dynain* file.

Each of these simulations took 24 hours to run with 120 cpus each. Once they were all simulated, everything was ready to extract and analyze the results.

Chapter 3

Analysis of Results

The analysis of the results was performed mainly in *META*, *Python* and *CORA*. *META* was used for checking the kinematics of the different simulations, and for data extraction related to the *SAFER A-HBM*. *Python* was used for data handling, and preparation of several curves and graphics to better understand the results obtained. Lastly, *CORA* was used for calculating the correlation between the different passive simulations with the corresponding active ones.

The active simulations will be treated as baselines to which the corresponding passive simulations will be compared. This way, the analysis of the results will be divided into four big different groups: the four pre-crash and in-crash combinations developed in this Project.

For each of these groups, excursions in the head (CoG[1], from node 4100011), chest (T4[2], from node 1990041) and pelvis (CoG, from node 1999999) will be extracted and compared graphically in *XZ-plane* and *XY-plane*. These excursions were taken relatively to the car position.

SAFER A-HBM data was extracted from each simulation to calculate the principal injury parameters and criteria. From these, the selected ones for this Project were, from up to bottom: HIC (Head Injury Criteria), BrIC (Brain Injury Criteria), highest brain strain value (at grey matter, white matter and corpus callosum), NIC (Neck injury criteria), spine compression loads and risk of rib fracture for a three different range of ages (25, 45 and 75 years old).

Results in this Project are confidential. Therefore, the ones presented in this section

[1]Center of Gravity.

[2]Thoracic vertebrae 4.

are normalized compared to the baseline, which means that all the injury criteria have value 1.0 for the baseline, and the rest of the simulations present increases and decreases compared to the baseline. For original results, please go to *Appendix* **??**.

3.1 Pre-crash: braking

The first presented results will be the ones obtained for the braking pre-crash maneuver. As shown in *Table 2.1*, this avoidance maneuver was expected to be followed by both a far side and a near side crash.

For the first load-case, the results will be commented and explained in a more detailed way in order to have a better understanding of the graphics and figures showed.

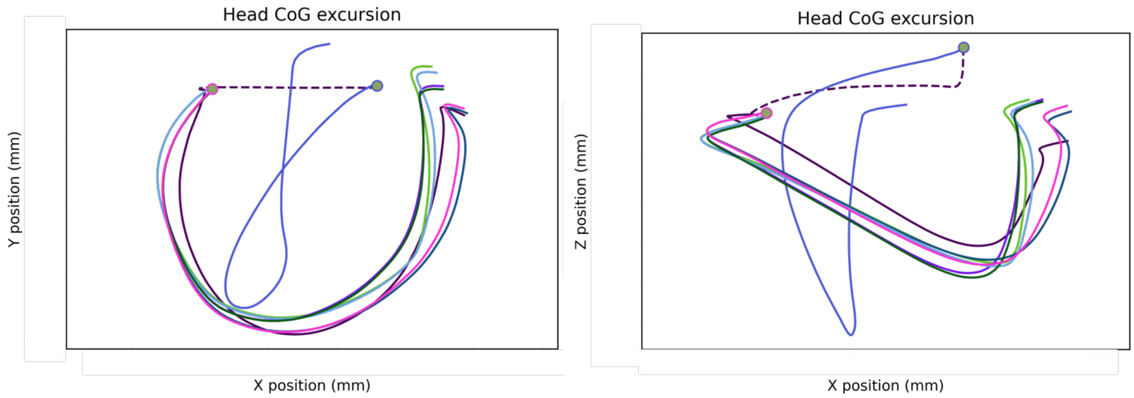
3.1.1 Braking - far side

This complete sequence consisted on a braking pre-crash maneuver followed by a far side in-crash scenario.

Head CoG, T4 and pelvis CoG excursions are presented below. In the following nomenclature, P stands for position, V for velocity, S for stresses and M for retained muscle activity level from the last muscles PID signal.

In these graphics, an excursion comparison between the baseline and each of its corresponding seven passive simulations is shown for the CoG of the head, T4 and CoG of the pelvis. Two points can be appreciated in the graphics. One of them indicates the start of the motion from the HBM nominal position (only in the case of the baseline simulation and the passive simulation with no retained parameters). On the other hand, the second point indicates the start of the motion when retaining the posture of the HBM (simulations 1 - 6).

Logically, the path that goes from the nominal position point to the retained position point (dahed line) corresponds to the pre-crash motion of the head, T4 or pelvis, respectively for each graphic. Another good appreciation would be that there is clearly one simulation very different from the rest, which corresponds with the passive simulation with no retained parameters. This simulation started from the nominal point and had the in-crash pulse directly applied. Thus, by not having even the pre-crash position retained, it had a completely different motion in comparison with the rest of the simulations.



(a) Head CoG excursions *XY-plane* (b) Head CoG excursions *XZ-plane*

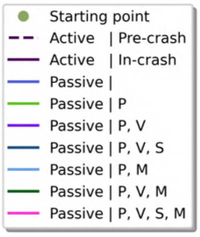
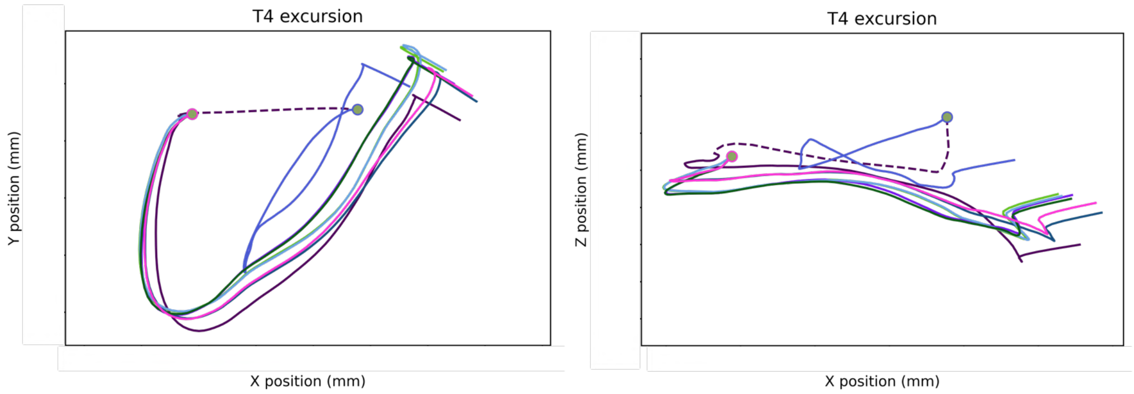
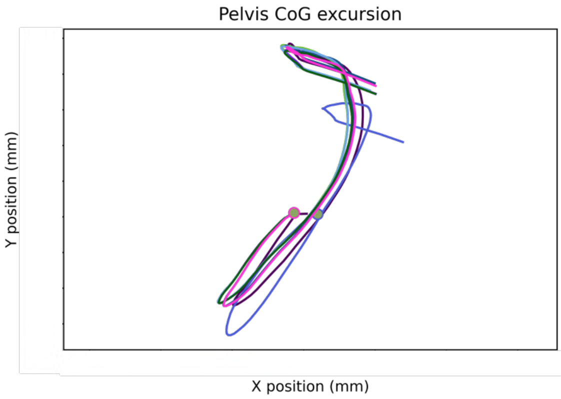


Figure 3.1: Head CoG excursions. Pre-crash: braking, in-crash: far side

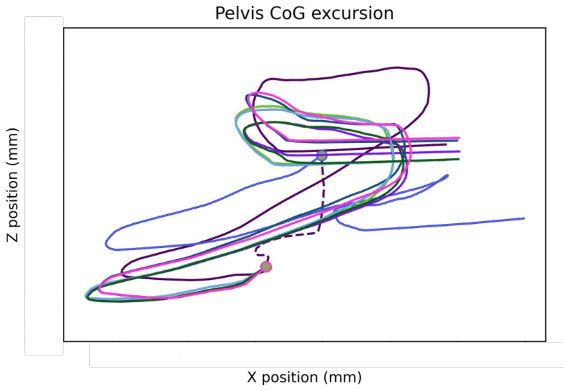


(a) T4 excursions *XY-plane* (b) T4 excursions *XZ-plane*

Figure 3.2: T4 excursions. Pre-crash: braking, in-crash: far side



(a) Pelvis CoG excursions *XY-plane*



(b) Pelvis CoG excursions *XZ-plane*

Figure 3.3: Pelvis CoG excursions. Pre-crash: braking, in-crash: far side

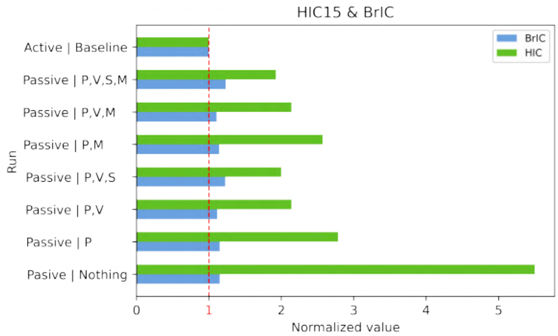
Another issue that can be seen in all of the *XZ-plane* graphics is that there is an initial vertical displacement of the head, T4 and pelvis in the active simulation. After further analysis, it was concluded that this was due to the fact that the HBM had not been squashed deep enough into the foam of the seat when preparing the models. So the 300 ms of initialization at the beginning of the simulation, which were supposed to be enough time for the HBM to reach equilibrium with the seat, were not enough. Therefore, the pre-crash started some miliseconds after the HBM could reach equilibrium with the seat, which could have partially affected the results.

Visually from the excursion graphics, there can also be seen that the passive simulation with no retained parameters is the most different to the baseline. However, when inputting the retained position, the excursions start to look much more similar to the baseline motion. Consequently, as more parameters are included into the simulations, it was expected that the kinematics would be even closer to the motion of the baseline. In this load-case, it can be appreciated that the two simulations with retained stresses are the closest ones to the baseline, what could confirm the fact that the more parameters added, the more similarities with the baseline simulation were obtained, as these two simulations were the ones with the most retained parameters.

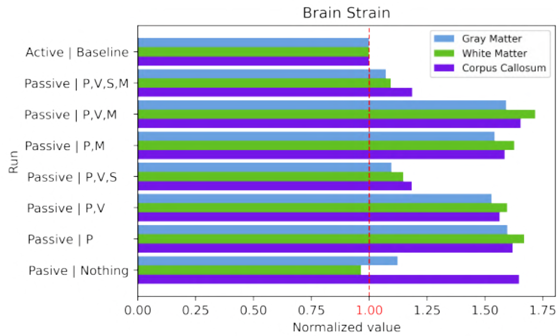
However, the excursions were still not the same. The simulation with all the parameters included, which can be considered as a passive equivalent for the baseline in-crash, had still differences with it. This is where the great difference between the baseline and passive simulations came in: active musculature. Which appeared to be the reason for these differences. Thus, regarding the principal objective of this Project, the active musculature appeared to have influence when comparing with a passive HBM with the same load-case applied.

However, part of these differences in excursions could also be due to the fact that the belt had to be re-routed and re-initialized, which means that the routing and the initial forces were not exactly the same as in the baseline. This will be mentioned in the discussion.

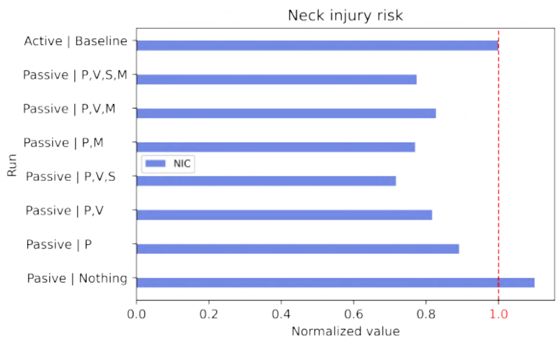
The graphs corresponding to the injury criteria extracted from the model are shown in *Figure 3.4*.



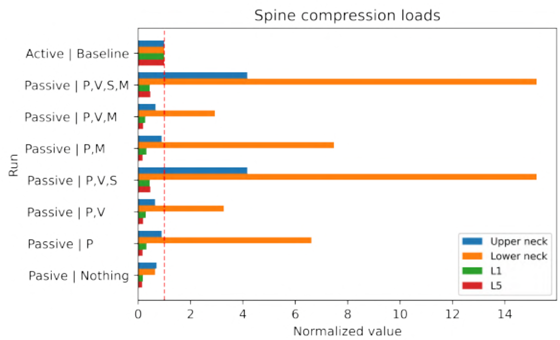
(a) HIC and BrIC. Normalized.



(b) Brain strain. Normalized.



(c) Neck injury risk. Normalized.



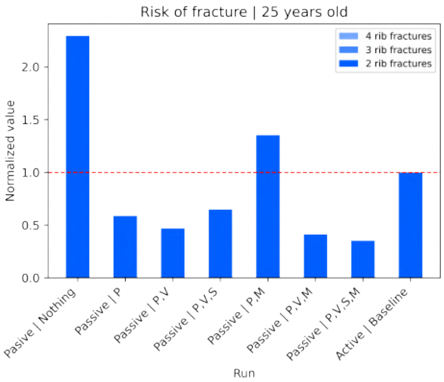
(d) Spine compression loads. Normalized.

Figure 3.4: Normalized injury criteria. Pre-crash: braking, in-crash: far side

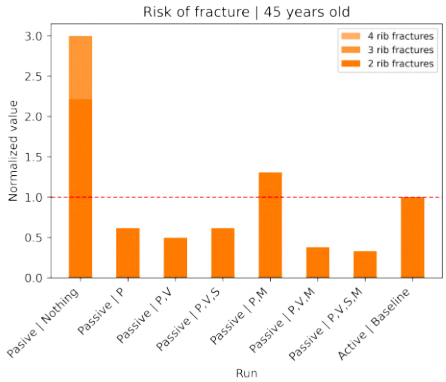
As it can be seen, HIC value in the passive simulations was always higher than the one predicted in the baseline. However, it decreased towards the baseline value as retained parameters were added, as expected. All the parameters appeared to have influence on the injury risk prediction when it came to the HIC. Besides, a tendency was detected. When accumulating retained parameters of posture, velocity and stress, HIC (and generally excursions) was closer to the baseline value. However, muscle stiffness obtained from the last PID signal (included in simulations 4 -

6) presented a very low influence, and sometimes zero influence when comparing to other passive simulations. This was found for the HIC and also for the excursions, where muscle stiffness caused some offset in the behaviour of the simulations. BrIC value was a little higher for the passive simulations than the predicted for the baseline, but it had no visible changes when retained parameters were included.

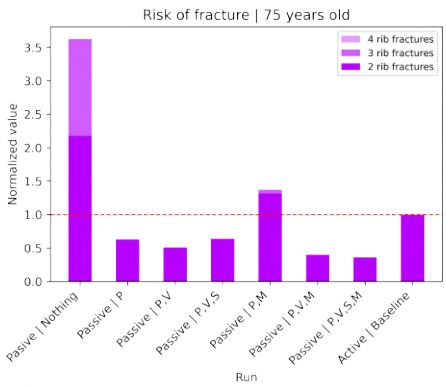
When talking about the rest of the injury criteria, it was found that the passive model usually over predicted the injury risk in comparison to the active model. Being very remarkable in the brain strain, upper neck and lower neck, having the last two a notable increase when including posture and/or stresses.



(a) Risk of rib fracture. 25 years old. Normalized.



(b) Risk of rib fracture. 45 years old. Normalized.



(c) Risk of rib fracture. 75 years old. Normalized.

Figure 3.5: Normalized rib risk fracture. Pre-crash: braking, in-crash: far side

The predicted rib fractures depending on three different ages is shown in *Figure 3.5*

Although the results are a little more complex, due to they show the prediction for two, three and four rib fractures, the general tendency, regardless of the age, is to underestimate rib fracture, except for the passive simulation with no retained parameters, which predicts much more risk than the baseline. This last detail was expected. However, there is a clear outlier, which is the retained muscle stiffness added to the position, which also over predicts the rib fracture.

It is remarkable to mention that, as explained, these results are normalized with the baseline, so a great increase or decrease in comparison to the baseline does not necessary mean that the real value has that much of a difference.

CORA correlations results are shown in *Figure 3.6*. These correlations were carried out based on the output accelerations from the head CoG, T4 vertebrae and pelvis CoG. The selected weights for the correlation analysis were 0.3 for *X-axis*, 0.5 for *Y-axis* and 0.2 for *Z-axis*. More weight was given to *Y-axis*, and secondly to *X-axis*, due to the in-crash load-case was principally lateral, while pre-crash was longitudinal. Correlation increased as retained parameters were included, reaching a value of 0.8 for simulation 3. However, muscle activity level retainment lowered the resuls, as it can be appreciated in the last three simulations.

These models turned out to be very sensitive when comparing accelerations between the passive simulations and the baseline, that is why *CORA* scores were not very high. Nevertheless, if correlation analysis would have been done based on excursions, results may have been higher for these simulations.

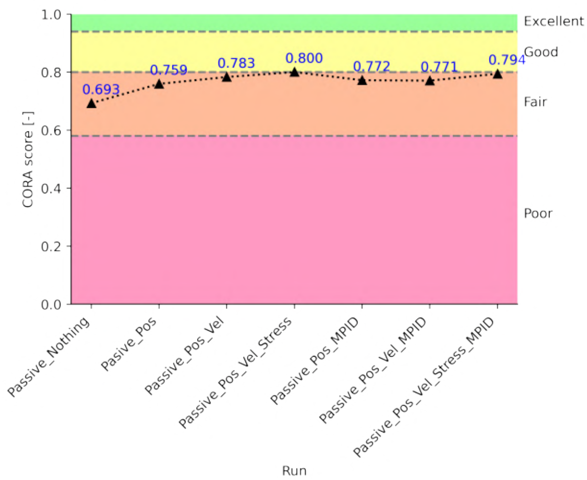
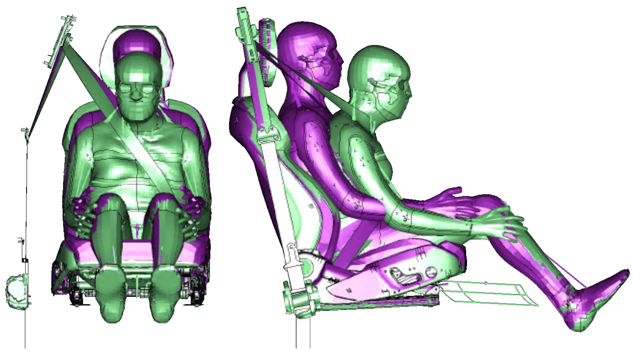


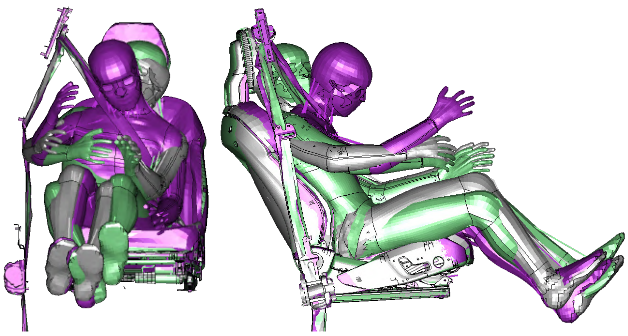
Figure 3.6: CORA score. Pre-crash: braking, in-crash: far side.

In *Figure 3.7*, a visual comparison is shown between three different simulations, which are the baseline[3] (grey), the passive simulation with no retained parameters (purple) and the passive simulation with all the retained parameters (green). As it can be seen, the complete passive simulation has a final state much more similar than the simple passive one (visible at *Figure 3.7c*). This is what was expected and is the reason to justify higher correlation values based on excursions.

However, during the baseline simulation, the model slided from the belt through the shoulder, which never happened for any of the passive simulations. This could be attributed to the re-routing of the belt when restarting the passive simulations, which could have partially affected the motions, as well as the injury prediction results and correlation values.



(a) Original state. Front view. (b) Original state. Side view.



(c) Final state. Front view. (d) Final state. Side view.

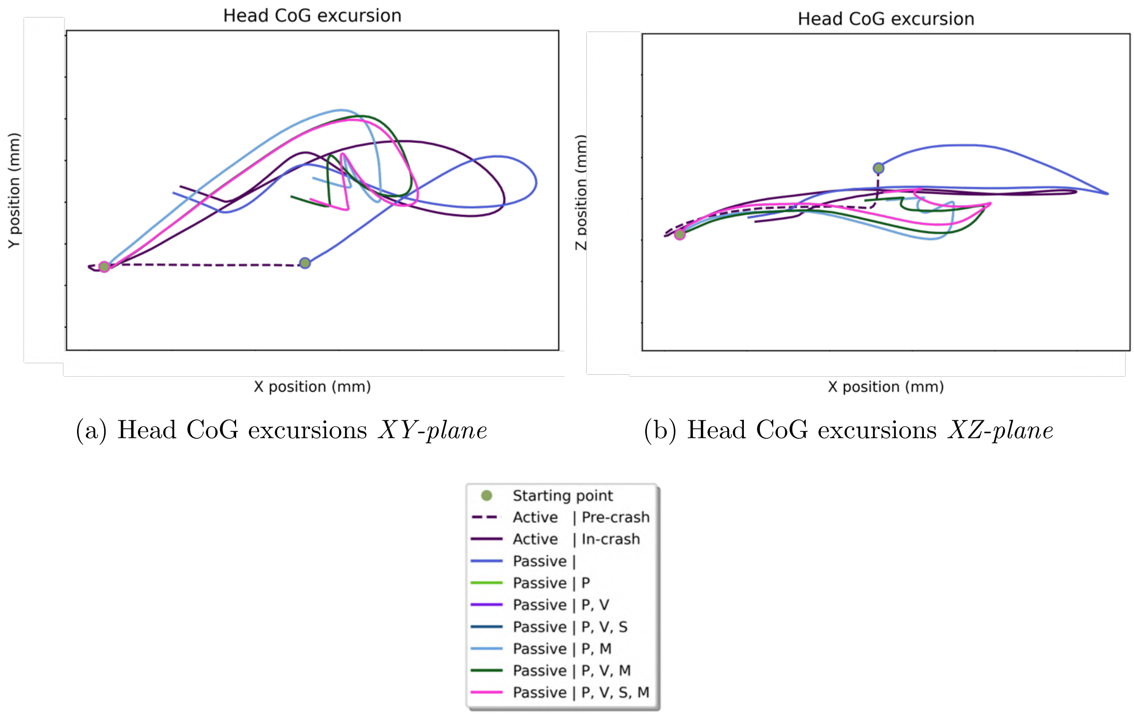
Figure 3.7: Original and Final state comparison. Grey: baseline, purple: no retained parameters, green: all retained parameters. Pre-crash: braking, in-crash: far side

[3]Baseline simulation (grey) is not visible in *Figures 3.7a and 3.7b* due to it has the same initial position as the complete passive simulation (green).

3.1.2 Braking - near side

This complete sequence consisted on a braking pre-crash maneuver followed by a near side in-crash scenario.

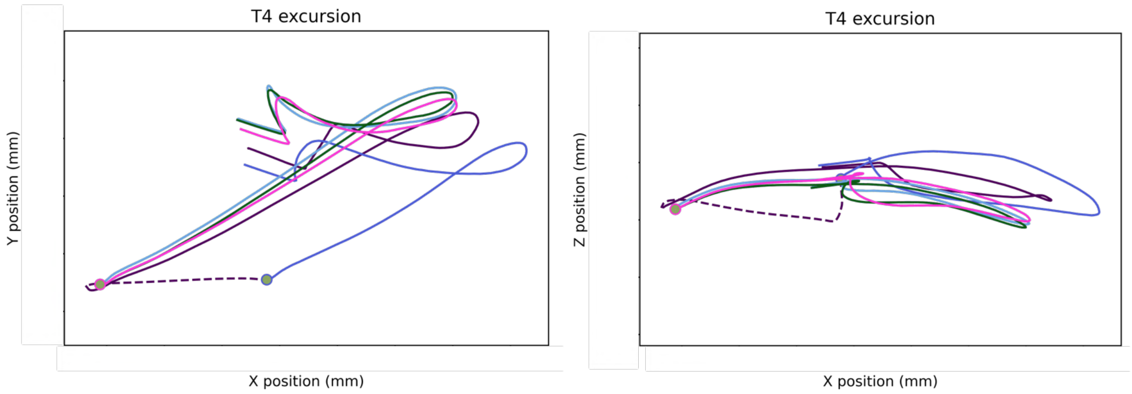
Observing the excursions, all the passive simulations have similar kinematics with each other. However, the final state of the passive with no retained parameters is very similar to the baseline when it comes to the head CoG and the T4 vertebrae. This turned out to be curious, due to this simulation starts from the nominal position point, and makes a larger displacement to end with a similar motion as the baseline. On the other hand, great differences between the rest of the passive simulations and the baseline can be attributed to the shoulder belt, which was re-routed for these simulations (not for the passive without retained parameters). The different routing combined with the fact that the initial stresses and forces from the belt could not be retained could explain this behaviour.



(a) Head CoG excursions *XY-plane*

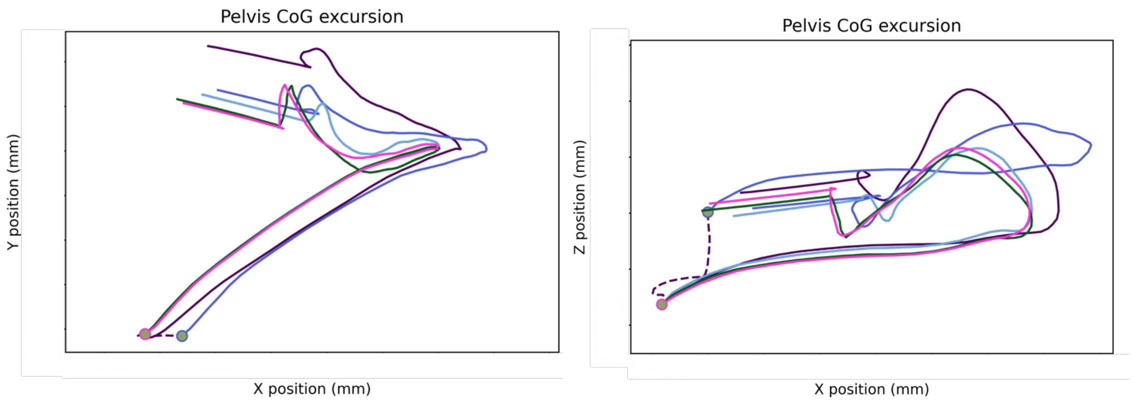
(b) Head CoG excursions *XZ-plane*

Figure 3.8: Head CoG excursions. Pre-crash: braking, in-crash: near side



(a) T4 excursions *XY-plane* (b) T4 excursions *XZ-plane*

Figure 3.9: T4 excursions. Pre-crash: braking, in-crash: near side

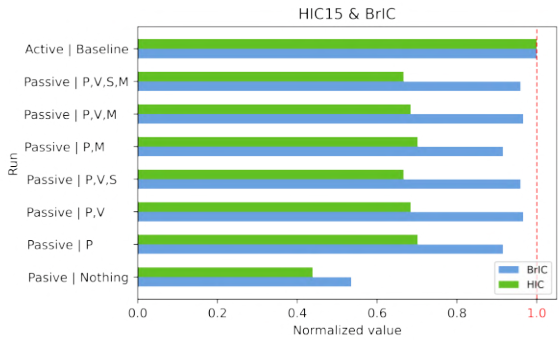


(a) Pelvis CoG excursions *XY-plane* (b) Pelvis CoG excursions *XZ-plane*

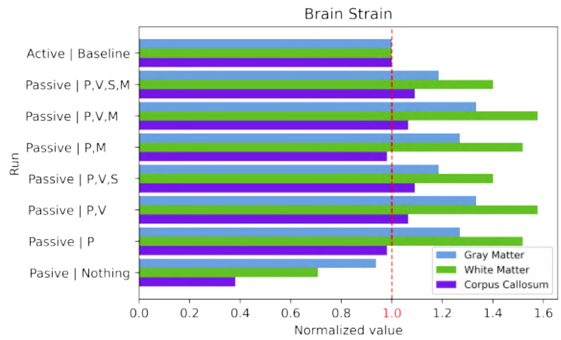
Figure 3.10: Pelvis CoG excursions. Pre-crash: braking, in-crash: near side

However, regarding the pelvis CoG, the passive simulation with no retained parameters stopped being the most similar one to the baseline. In this particular case, the pelvis CoG excursion was very affected by the squashing with the seat, which had not yet reached equilibrium in the nominal position. This can be seen in *Figure 3.10b*.

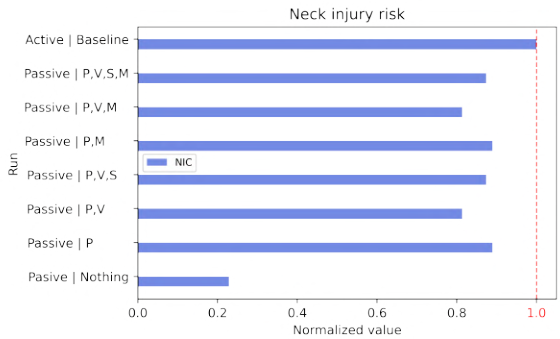
In *Figure 3.11* injury criteria extracted from the model are presented. Some different and interesting outcomes were found for this load-case.



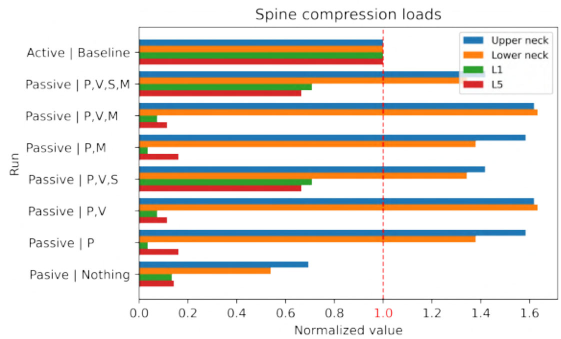
(a) HIC and BrIC. Normalized.



(b) Brain strain. Normalized.



(c) Neck injury risk. Normalized.

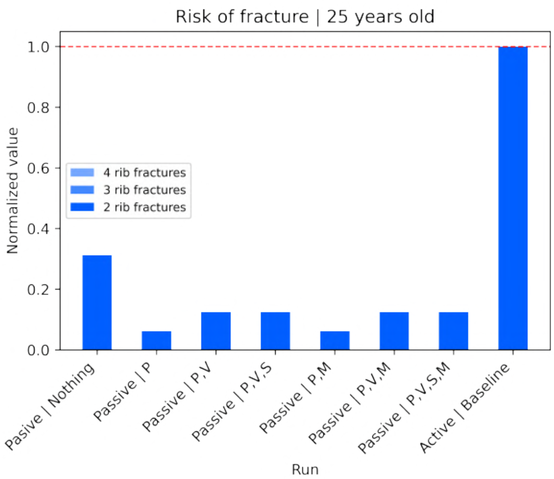


(d) Spine compression loads. Normalized.

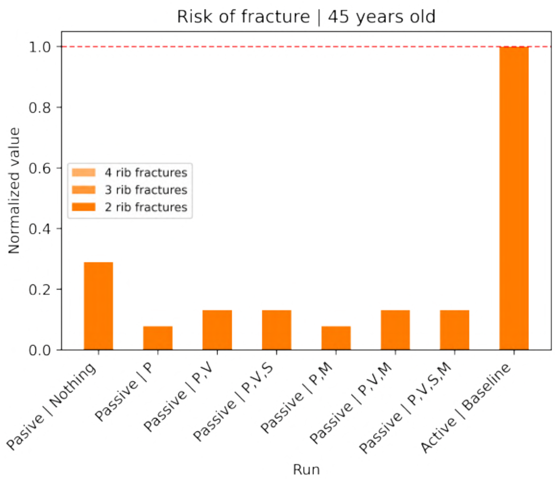
Figure 3.11: Normalized injury criteria. Pre-crash: braking, in-crash: near side

For this near side load-case, HIC and BrIC values had a lower injury prediction for all the passive cases, which was a completely opposite behaviour than the one with the previous load-case, under predicting the injury risk in comparison with the baseline. Accelerations of the Head CoG were lower for this load-case, as the head had quite an early contact with the interior curtain airbag. The passive simulation with no retained parameters was the one with the lowest HIC value, as the head CoG had a smoother slope in X acceleration at the time of the contact with the airbag compared with the rest of the simulations. This made sense because when retaining the position of the HBM in a braking maneuver, the principal component altered is the *X-axis*, right where the major difference was found. Following the same behaviour, BrIC and NIC values were also under predicted, without remarkable changes except for the simulation with no retained parameters, which was even lower. This was thought to be due to the same reason as for the HIC value commented above. Thus, retaining the posture appeared to have a great influence when it comes to injury prediction.

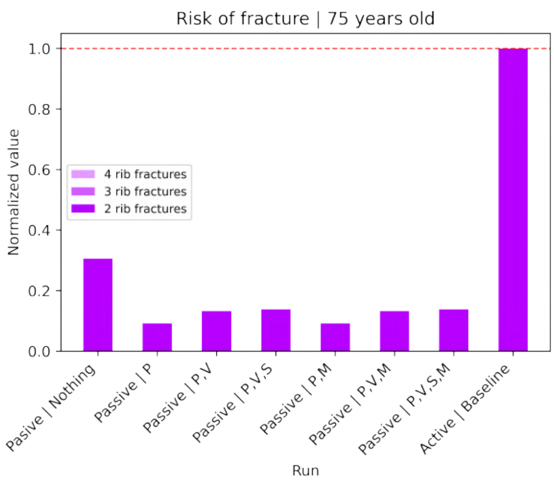
The brain strain was, by contrast, over predicted by every passive simulation except for the passive with no retained parameters. Same behaviour was found for the upper and lower neck. L1 and L5 had an increase when including stresses, but still under predicting.



(a) Risk of rib fracture. 25 years old. Normalized.



(b) Risk of rib fracture. 45 years old. Normalized.



(c) Risk of rib fracture. 75 years old. Normalized.

Figure 3.12: Normalized rib risk fracture. Pre-crash: braking, in-crash: near side

In the case of the rib fracture prediction, the model had the same normalized prediction for the three analyzed ages. Contrary to the previous load-case, with the near side crash all the simulations under predicted the risk of rib fracture when comparing it to the baseline model. The re-routing of the belt could also have affected these results, by starting from a re-routed belt with no retained stresses and forces, which could be compressing the rib cage less from the beginning.

It is important to keep in mind that these results are normalized with the baseline, which presented very low values for all injury risks. Thus, the relative values for the passive simulations do not differ that much from the baseline. The problem comes when dividing a value by a number very close to zero, as the result would be very sensitive to small changes.

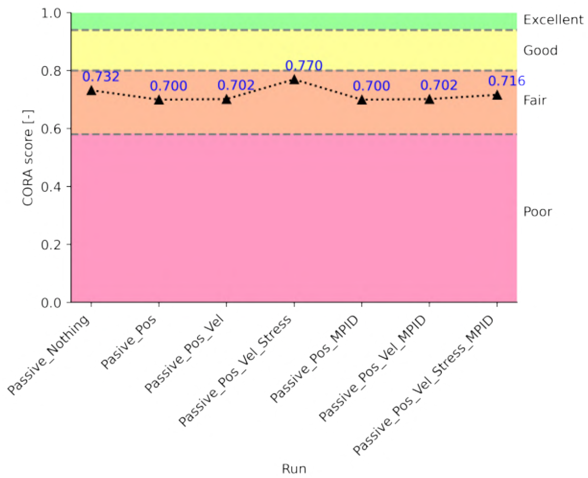
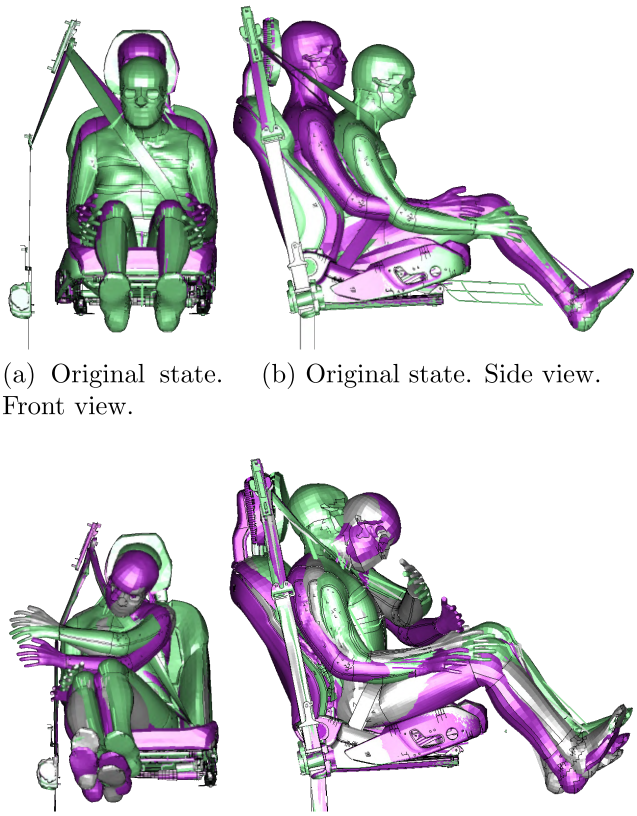


Figure 3.13: CORA score. Pre-crash: braking, in-crash: near side.

Regarding the correlation values from *CORA*, which were based on accelerations, results do not have remarkable changes, being the simulation with the highest correlation with the baseline the one with posture, velocity and stresses included. Once again, the retained muscle activity level did not seem to provide a good correlation for the model.

This *CORA* results were obtained applying the same weights as in the previous load-case: 0.3 for *X-axis*, 0.5 for *Y-axis* and 0.2 for *Z-axis*.

Looking at the final position of the HBM in *Figure 3.14* at the end of the passive simulation with no retained parameters (purple) and the passive simulation with all the retained parameters (green) compared to the baseline (grey), it can be seen that it is the purple HBM the one that has a closer posture to the baseline HBM. This was a strange result to have in this load-case. However, it should also be taken into account that this is not a near side crash, but a far side crash flipped to try to replicate a near side scenario. Besides, intrusions, which is a parameter to be taken into account for near side crashes [20], were not included in this Project due to a matter of time. Therefore, the results obtained for this load-case are considered as a study, but are not specifically representative of a real near side crash.



(a) Original state. Front view.

(b) Original state. Side view.

(c) Final state. Front view.

(d) Final state. Side view.

Figure 3.14: Original and Final state comparison. Grey: baseline, purple: no retained parameters, green: all retained parameters. Pre-crash: braking, in-crash: near side

3.2 Pre-crash: left lane change

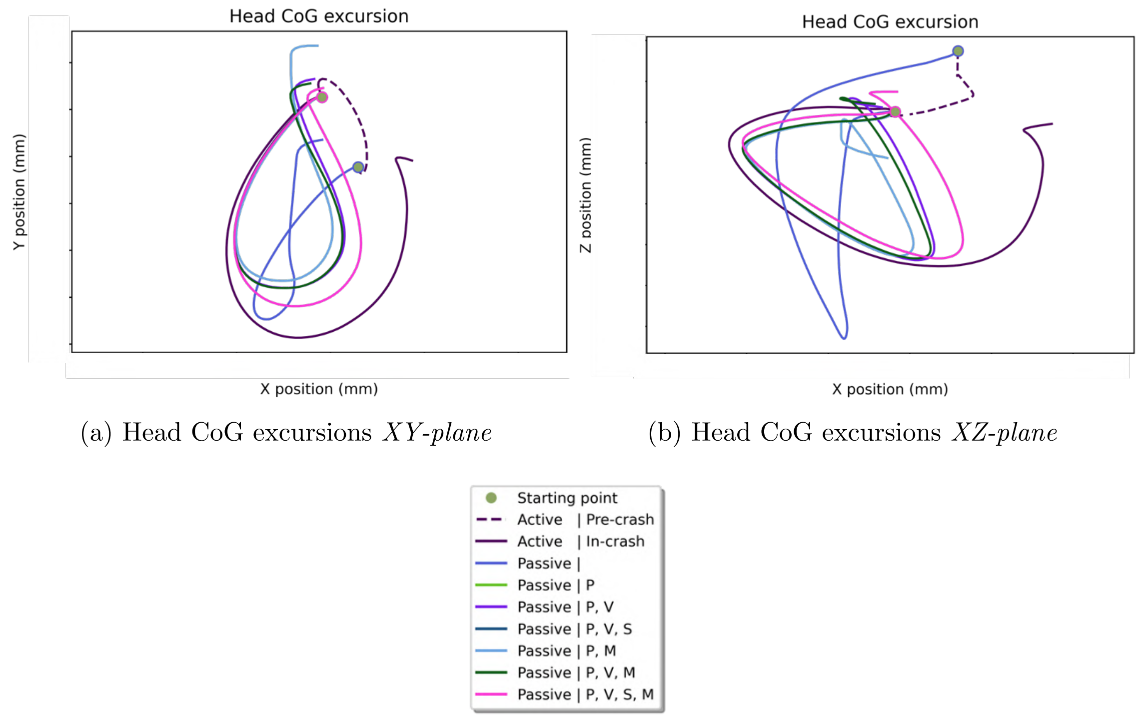
The next pre-crash to analyze was the left lane change, which according to *Table 2.1*, was expected to be followed by a far side crash.

3.2.1 Left lane change - far side

This complete sequence consisted on a left lane change pre-crash maneuver followed by a far side in-crash scenario.

From excursions, it can be seen that be appreciated that in the case of the head CoG, the passive simulation with all the retained parameters included was the closest one to the baseline. Looking at the passive simulation with no retained parameters included, a completely different motion could be seen. This is due to the fact that, as not even the position was retained, the head collided with the central armrest, having a rebound afterwards. It is because of this that a higher HIC value was expected for this specific simulation.

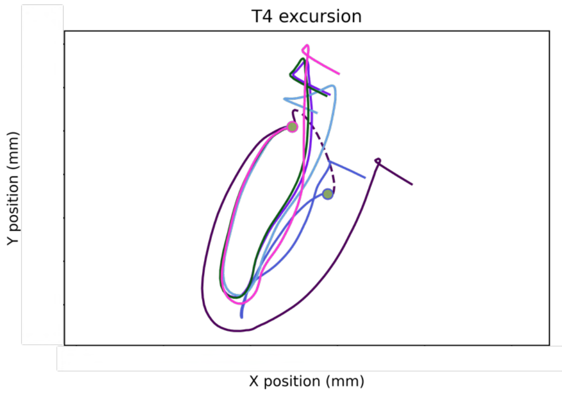
However, there was not that much of a difference between the passive simulations, including the one with no retained parameters included, for the T4 vertebrae and the pelvis CoG.



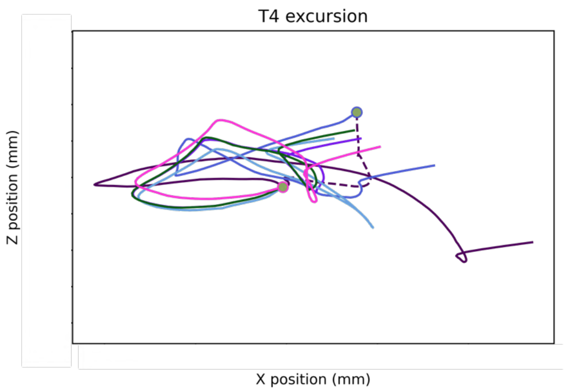
(a) Head CoG excursions *XY-plane*

(b) Head CoG excursions *XZ-plane*

Figure 3.15: Head CoG excursions. Pre-crash: left lane change, in-crash: far side

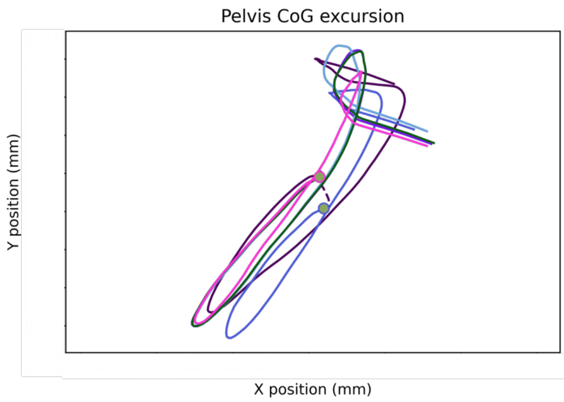


(a) T4 excursions *XY-plane*

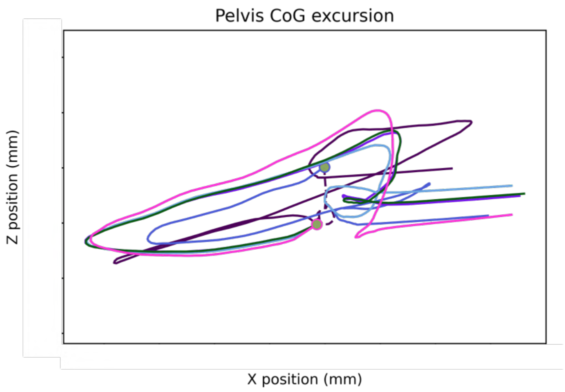


(b) T4 excursions *XZ-plane*

Figure 3.16: T4 excursions. Pre-crash: left lane change, in-crash: far side



(a) Pelvis CoG excursions *XY-plane*



(b) Pelvis CoG excursions *XZ-plane*

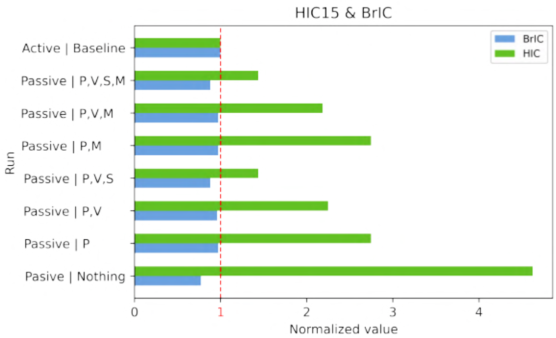
Figure 3.17: Pelvis CoG excursions. Pre-crash: left lane change, in-crash: far side

In *Figure 3.17b*, the issue with the squashing of the seat can be appreciated, as both the baseline and the passive simulation with no retained parameters started the motion going downwards, while the rest of the passive simulations, that had already reached equilibrium during the pre-crash, had a different motion, going upwards.

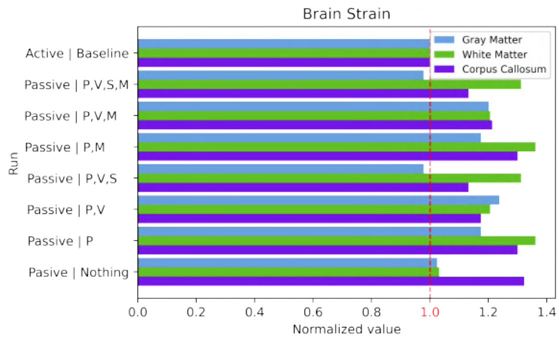
Taking a look at the injury risk criteria extracted from the models shown in *Figure 3.18*, a similar behaviour as in the first load-case of this Project (braking pre-crash followed by a far side scenario) was found. HIC value was over predicted by the passive simulations when comparing with the baseline. Besides, the passive simulation with no retained parameters had the largest HIC, as expected from the collision of the head with the central armrest.

HIC got closer to the baseline prediction as long as more retained parameters were included, finding no visible influence for the retained muscle activity level. But still, all the values were over predicting when comparing to the baseline.

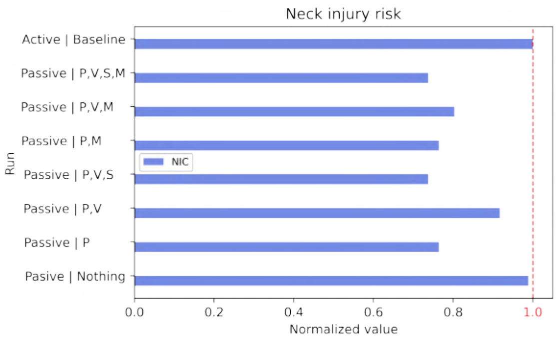
BrIc and brain strain estimations had the same behaviour as in the first far side scenario analyzed in the Project. However,lower neck loads were over predicted by the passive simulations.



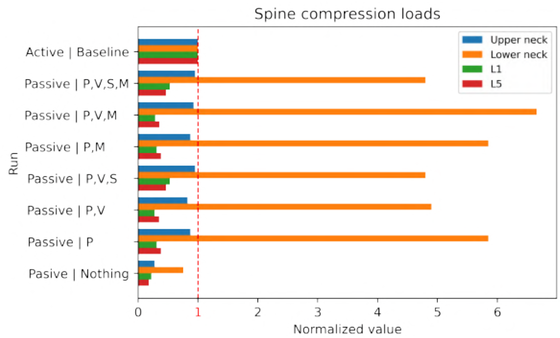
(a) HIC and BrIC. Normalized.



(b) Brain strain. Normalized.



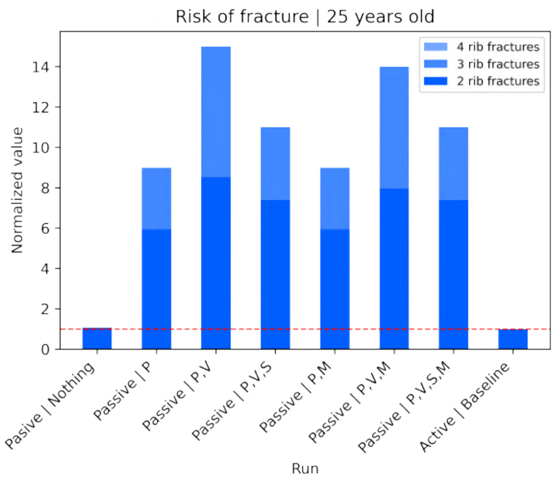
(c) Neck injury risk. Normalized.



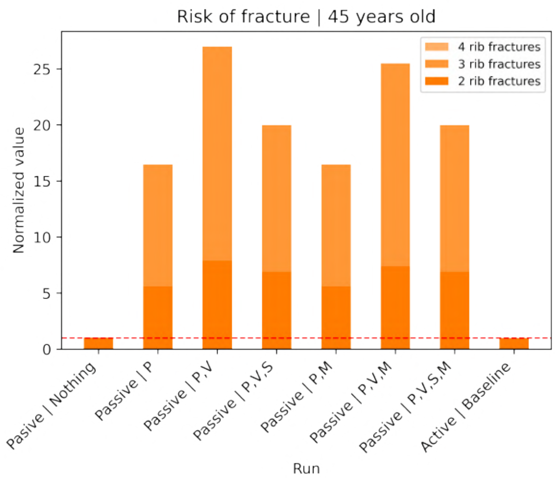
(d) Spine compression loads. Normalized.

Figure 3.18: Normalized injury criteria. Pre-crash: left lane change, in-crash: far side

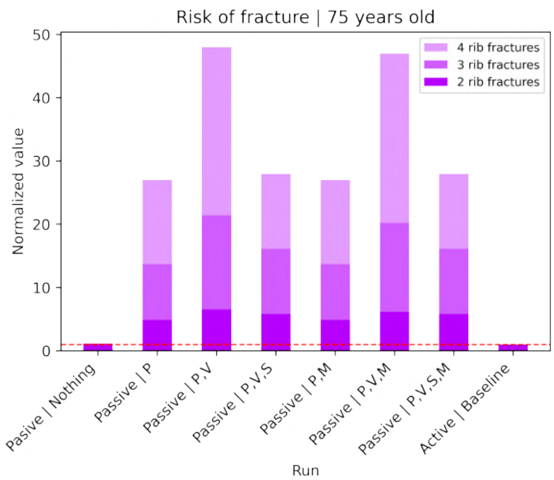
In *Figure 3.19*, risk of rib fractures are shown, which again, had the same relative values with the baseline for every age analyzed in this Project. All passive simulations were over predicting compared to the baseline.



(a) Risk of rib fracture. 25 years old. Normalized.



(b) Risk of rib fracture. 45 years old. Normalized.



(c) Risk of rib fracture. 75 years old. Normalized.

Figure 3.19: Normalized rib risk fracture. Pre-crash: left lane change, in-crash: far side

In this load-case, injury predictions were higher than in the near side, especially the risk of rib fracture. This could be so due to the fact that a far side load scenario is more aggressive to the rib cage, as the seatbelt becomes an important load when retaining the passenger due to the way it is designed.

However, increasing values showed in *Figure 3.19* are still not representative for the real values, as they were still low.

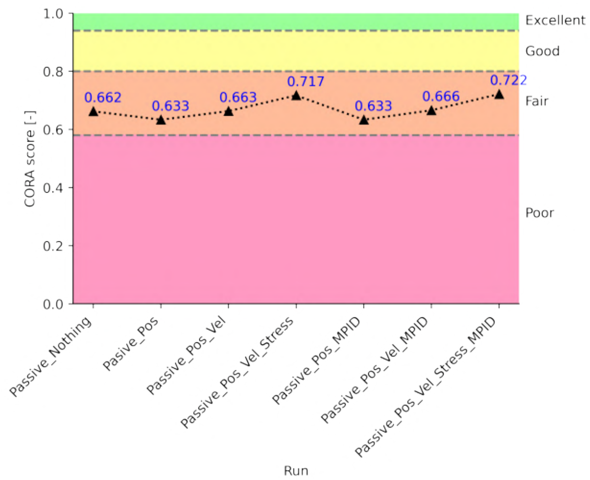


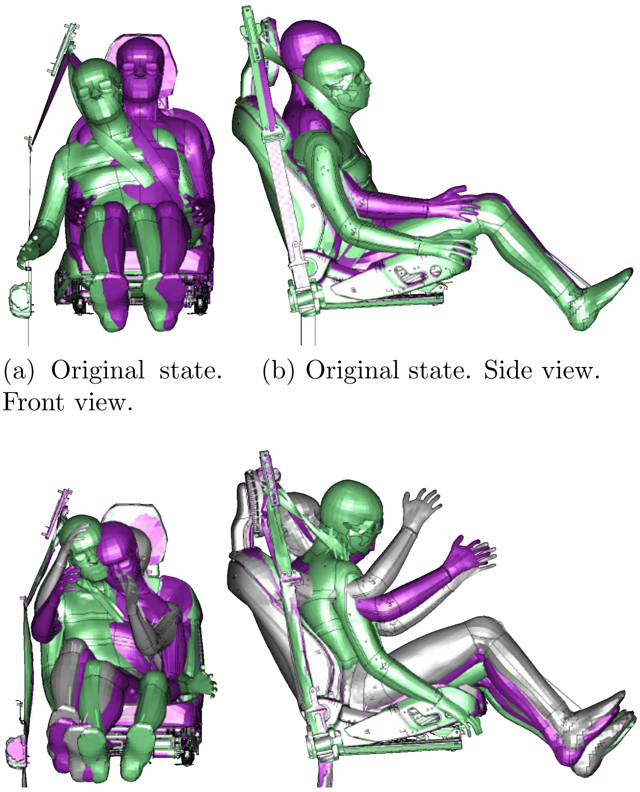
Figure 3.20: CORA score. Pre-crash: left lane change, in-crash: far side.

Correlation values from *CORA*, based on accelerations, are shown in *Figure 3.20*. In this case, as in the previous one, weights were 0.3 for *X-axis*, 0.5 for *Y-axis* and 0.2 for *Z-axis*.

In this case, as in the previous ones, the correlation values remained quite constant, getting the highest correlation for the passive simulation with every retained parameter included. Posture appeared to have a negative impact in correlation, which is thought to be due to the re-routing of the belt, affecting the initial kinematics of the simulation as initial stresses and forces were not retained from the pre-crash. If this would have been done, much higher correlation results would have been expected.

Lastly, taking a look at *Figure 3.21*, it can be appreciated than the purple HBM (passive with no retained parameters) was closer in position to the baseline than the green HBM (passive with every retained parameters). However, the final position of the purple HBM was not representative of its whole motion, as it reached that posture after hitting the central armrest.

In addition, in *Figure 3.21c*, a really interesting outcome can be appreciated. Focusing on the head, the difference between a passive, passive with retained muscle activity level and active model can be seen. The active HBM (grey) is the one that was able to keep its head more straight (due to the APF controllers). The passive HBM (purple), on the other hand, was the one that ended up with the head more down. In a middle point between these two, was the passive HBM with the constant muscle stiffness retained from the last PID signal from the pre-crash (green), which was able to keep the head straighten to some extend.



(a) Original state. Front view.
(b) Original state. Side view.
(c) Final state. Front view.
(d) Final state. Side view.

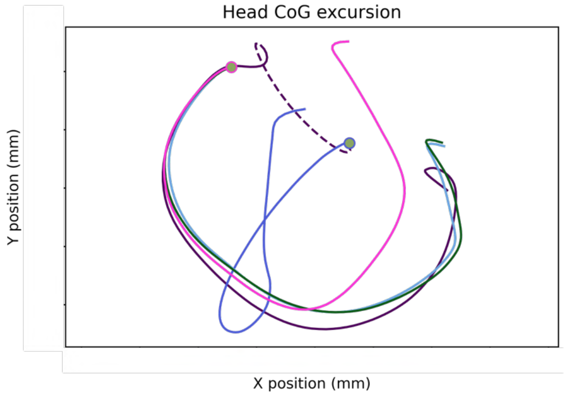
Figure 3.21: Original and Final state comparison. Grey: baseline, purple: no retained parameters, green: all retained parameters. Pre-crash: left lane change, in-crash: far side

3.3 Pre-crash: left lane change and braking

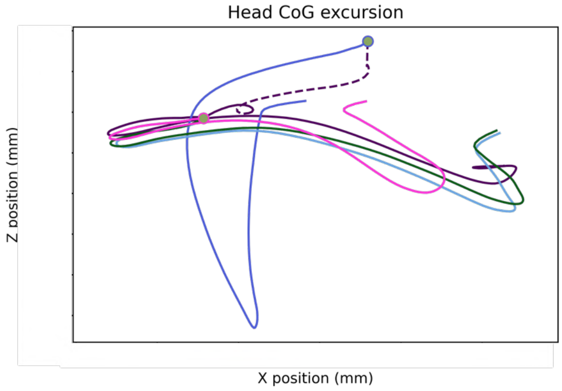
The last pre-crash to analyze was the left lane change, which according to Table2.1, was expected to be also followed by a far side crash.

3.3.1 Left lane change and braking - far side

This complete sequence consisted on a left lane change and braking pre-crash maneuver followed by a far side in-crash scenario.



(a) Head CoG excursions *XY-plane*



(b) Head CoG excursions *XZ-plane*

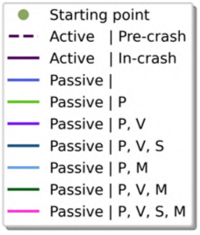
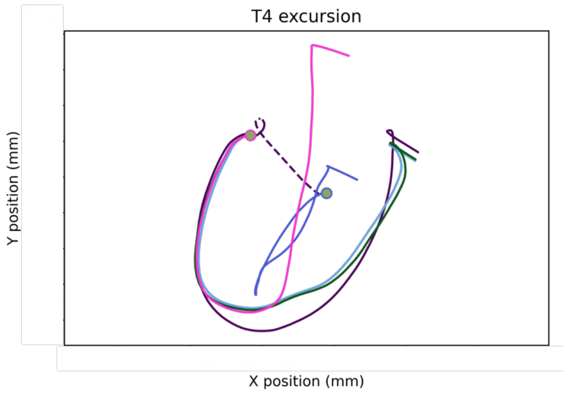
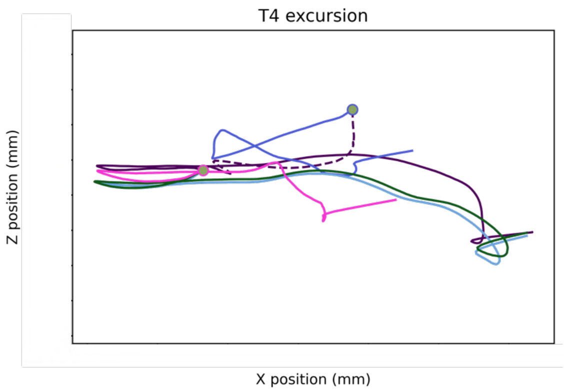


Figure 3.22: Head CoG excursions. Pre-crash: left lane change, in-crash: far side

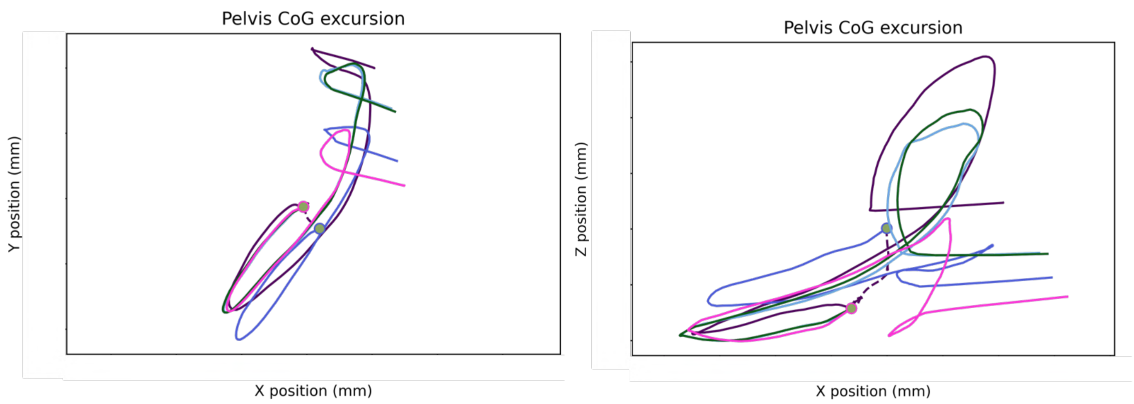


(a) T4 excursions *XY-plane*



(b) T4 excursions *XZ-plane*

Figure 3.23: T4 excursions. Pre-crash: left lane change, in-crash: far side



(a) Pelvis CoG excursions *XY-plane* (b) Pelvis CoG excursions *XZ-plane*

Figure 3.24: Pelvis CoG excursions. Pre-crash: left lane change and braking, in-crash: far side

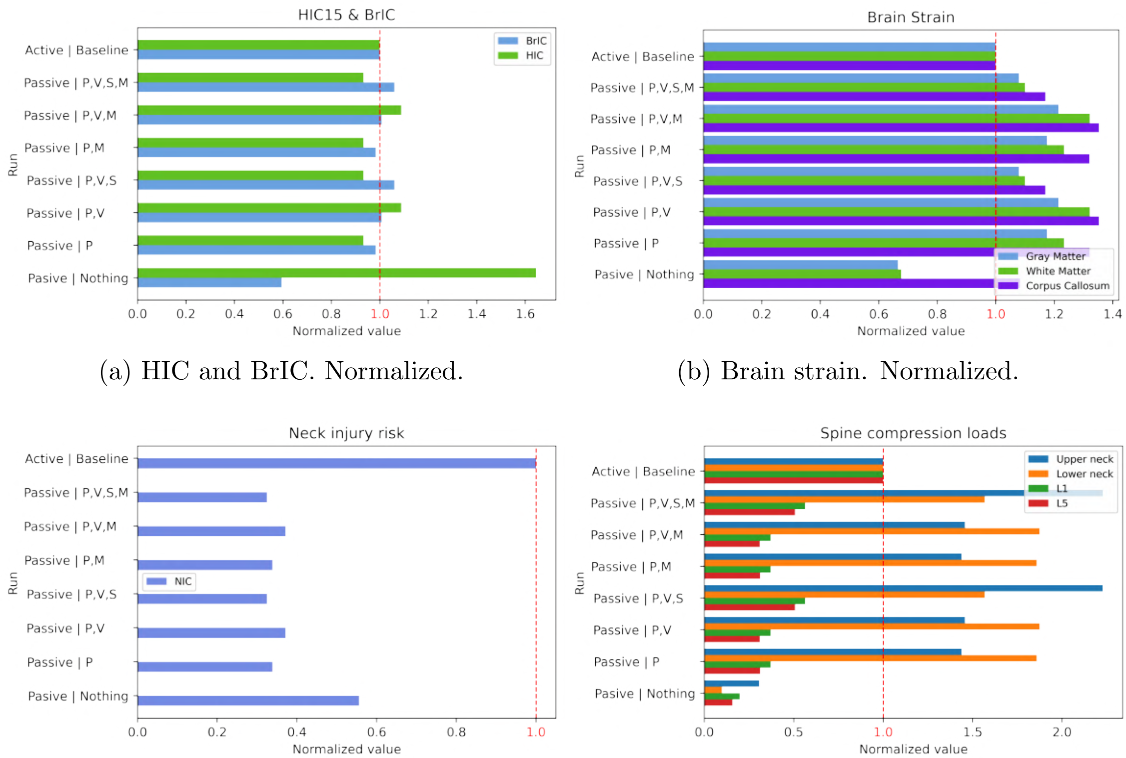
In this load-case, it can be appreciated that the passive simulation with all the retained parameters included was the one that differed the most with the baseline. This was a strange result that needed further investigation.

After examining the simulations, it was noticed that the baseline model did not have any slide of the belt at the shoulder. All of the passive models, on the other hand, did have slide of the belt at the shoulder, except for the ones that included the stresses. However, the kinematics were not the same as in the baseline, due to in these passive simulations with stresses included, the belt interacted with the clavicle in a rather abrupt way. This made the HBM to have a different motion compared with the rest of the passive simulations, and can be the reason why excursions with included stresses looked that different.

However, the fact that the passive simulations with stresses included were the ones that could better replicate the shoulder belt behaviour, was an interesting founding. Due to this belt behaviour, higher rib cage loads could be expected for passive simulations including stresses.

In addition, the passive simulation with no retained parameters had a very similar collision with the central armrest, as in the previous load-case. Thus, high HIC was expected for this simulation too.

In *Figures 3.22, 3.23 and 3.24*, and in some of the previous load-cases, it may look like there are some excursion curves missing. However, they are not. The issue is that when they were plotted, some of them had the exact same excursions as other passive simulations, so they were plotted one on top of each other. When this happened, it meant that there is a parameter that was not having any influence at all when being included, which only happened for the retained muscle activity level. Therefore, although *CORA* values are based in accelerations, same or very similar correlation values were expected when this happened. In this particular case, passive simulations 1 - 3 were not visible at the plots, which meant that the muscle retained muscle activity level had no influence at all in this load-case.



(a) HIC and BrIC. Normalized.

(b) Brain strain. Normalized.

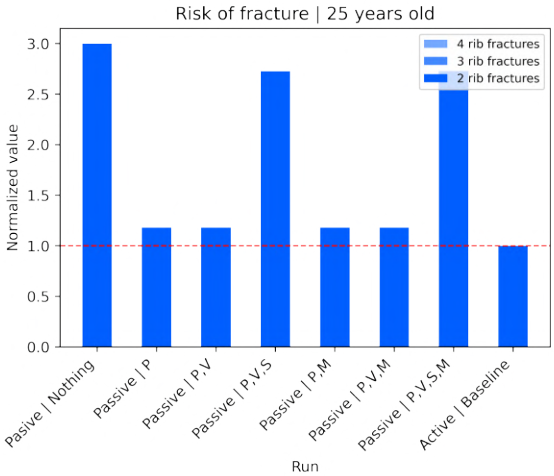
(c) Neck injury risk. Normalized.

(d) Spine compression loads. Normalized.

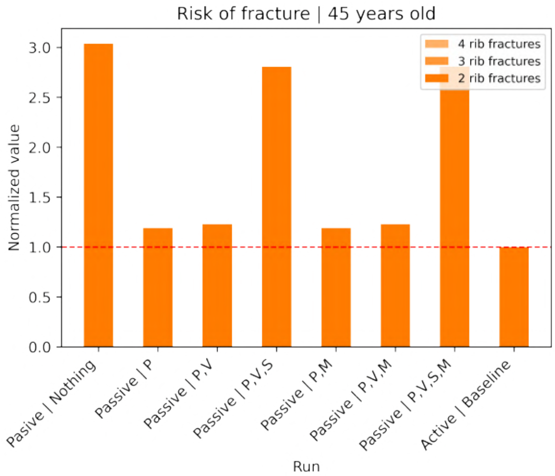
Figure 3.25: Normalized injury criteria. Pre-crash: left lane change and braking, in-crash: far side

As expected, HIC had a higher value, maybe due to the head impact with the central armrest. However, the rest of the injury predictions for the head were approximatedly the same as in the baseline.

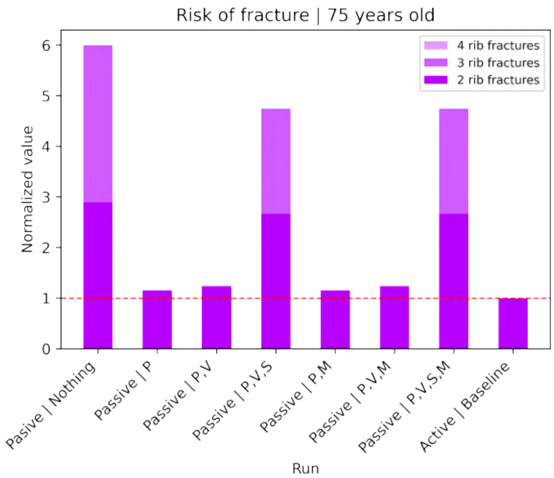
Spine compressions at L1 and L5, and NIC, were under predicted compared with the active model. On the other hand, upper and lower neck compression were over predicted for all the passive simulations with some retained parameter.



(a) Risk of rib fracture. 25 years old. Normalized.



(b) Risk of rib fracture. 45 years old. Normalized.



(c) Risk of rib fracture. 75 years old. Normalized.

Figure 3.26: Normalized rib risk fracture. Pre-crash: left lane change and braking, in-crash: far side

As expected from the belt behaviour mentioned above, higher rib fracture risk was predicted by the passive models with stresses included, as well as for the passive simulation with no retained parameters. For passive simulations with stresses included, also higher upper neck compression (see *Figure 3.25*) loads were predicted, this could be attributed to the fact that those simulations had also abrupt accelerations in the neck due to the belt interaction with the shoulder.

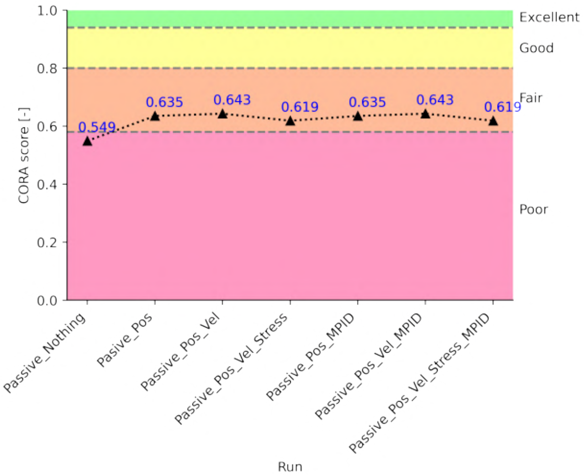


Figure 3.27: CORA score. Pre-crash: left lane change and braking, in-crash: far side.

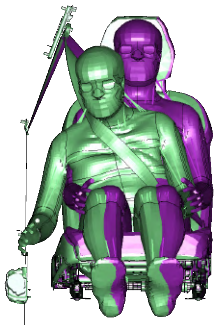
CORA correlation results based on accelerations (shown in *Figure 3.27*) show a poor correlation when simulating the passive HBM with no retained parameters. Correlation improves when including position and velocity, but it decreases when including stresses.

This decrease in correlation when including stresses was due to the shoulder belt behaviour, as the motion was completely different from the baseline. Moreover, it can be appreciated that simulations 1 - 3 had exactly the same correlation values as simulations 4 - 6. This fact justifies the overlapping of the excursions curves mentioned above[4].

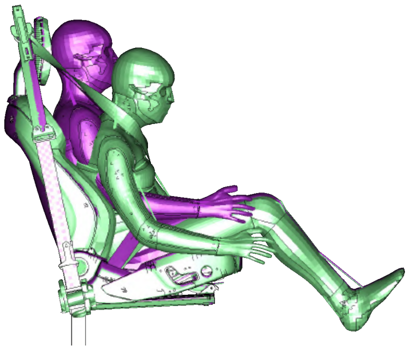
Lastly, taking a look at *Figure 3.28*, great differences can be spotted at the final state. The baseline HBM (grey) had a final state completely different from the rest of the HBMs. It is more similar to the passive HBM with no retained parameters (purple) (but still not close enough due to one of them had slide of the belt and the other did not), but completely different from the HBM with all the retained parameters, due to the belt compression in the shoulder.

Al these issues with the belt were again believed to be most likely related with the belt re-routing for the passive simulations, which turned out to be one of the major limitations of this project.

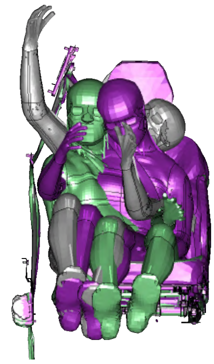
[4]As *CORA* calculations were based on accelerations, even if these correlation values were not identical, excursions may still have perfect overlapping, as it occured in previous load-cases.



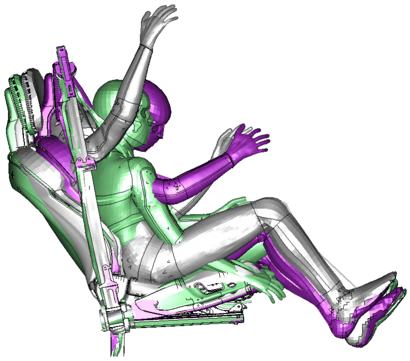
(a) Original state. Front view.



(b) Original state. Side view.



(c) Final state. Front view.



(d) Final state. Side view.

Figure 3.28: Original and Final state comparison. Grey: baseline, purple: no retained parameters, green: all retained parameters. Pre-crash: left lane change and braking, in-crash: far side

Chapter 4

Conclusions

This Project aimed to run several simulations utilizing the *SAFER A-HBM v10.0* with and without active musculature throughout different pre-crash and in-crash scenarios in order to study the influence of the active musculature on the injury risk prediction. Comparisons were made between the so-called active and passive simulations, which corresponded to the HBM with the active musculature turned on and off respectively.

Active simulations run first, simulating the whole pre-crash and in-crash sequence. Last state of the pre-crash was extracted from these simulations in order to re-run the in-crash with the HBM as passive including different retained parameters from the active pre-crash.

Results concluded that, generally speaking, the more retained parameters added from a passive in-crash to passive simulations, the closer the passive simulations were to the active ones. Retained muscle activity level from pre-crash turned out to maintain, and sometimes lower, the correlation results with active simulations. Besides, it had very low influence regarding injury risk predictions. Thus, results in this Project showed that this retained parameter may not be a useful one to be used in order to replicate an active model.

This suggested that muscles do not have influence during the in-crash[1], as the retained muscle activity level had low or zero influence when included into the passive simulations. This was expected, as the force developed by the muscles could be considered as negligible in comparison to the forces involved in the event of an in-crash [21].

[1]On the other hand, in [15] it was concluded that muscles turned out to have influence during the pre-crash maneuvers, i.e., low acceleration load-cases.

However, position was found out to be the retained parameter that had the greatest influence. In addition to this parameter, velocity and stresses also increased the correlation. Therefore, based on the results of this Project, if an active model wants to be replicated by a passive model, initial posture is the key parameter to include, with retained velocity and stresses being good additions for improving results if available.

Many differences were found between the most correlating passive model and the active one. Injury prediction was not accurately predicted by the passive models, especially by the ones with less retained parameters. However, even the most complete passive simulation had notorious deviations from the active model. Spine compression loads and neck injury risk predictors were giving bad results, with a generic over predicting tendency for upper and lower neck compression loads, and under predicted tendency for NIC and L1 and L5 compression loads.

This was attributed to the fact that no complete passive model that could adequately replicate the baseline model in-crash was simulated, mainly because of the seat and seatbelt extraction, which caused different kinematics in the occupant and, thus, different injury outcome.

Therefore, as a final comment, if an active model should be replicated from a passive model, the key parameter included in the model should be posture of the HBM. Velocities and stresses also showed improved results when adding them to the simulations. Retained muscle activity level did not improve correlation, proving that muscle activity does not influence the injury prediction during an in-crash load-case.

4.1 Limitations

There were some limitations encountered during the development of this Project. It should be taken into account that some of these limitations could have affected the results, and therefore may have influenced the conclusions drawn from them.

The first limitation of this Project was that there was not a complete passive simulation that could be used to fully compare it to the active model. The most accurate methodology would have been running the whole active pre-crash and deactivating active musculature when entering the in-crash scenario. This was not done mainly due to a matter of computational time, as it would have required around seven days per passive simulation. Instead, the passive model with posture, velocities, stresses and retained muscle activity level was considered a good approximation to the complete passive model.

A very important limitation to this project was the belt. Through the *Dynain* file extracted at the end of the pre-crash, it was possible to capture the state of the full HBM and the seat. However, seatbelt stresses and forces could not be included when restarting the passive simulations. Besides, due to the belt mechanisms could not be re-initialized, the belt had to be completely re-routed for the new posture in order to being able to run the passive in-crash simulations. This re-routing of the belt caused some sliding issues with the shoulder of the occupant in some of the passive simulations, which did not appear in the corresponding active simulations.

In addition to this, there was the issue with the seat squashing. The HBM was not pushed deep enough into the foam of the seat, so during the first 300 ms of initialization for the active simulations, the model did not have time to reach equilibrium between the HBM and the foam. So the pre-crash started before reaching that equilibrium. Besides, stresses of the structure of the seat were not included when restarting the passive simulations, which could have altered the kinematics of the occupant.

Both the issue with the belt and the seat squashing could have affected the results, and are probably the reason why it was not possible to recreate the baseline model with greater accuracy.

Regarding the near side crash, intrusions were not included, as the pulse came from a flipped far side crash. A great part of the possible injuries for the occupant during a near crash side are caused by intrusions, so the results obtained for this scenario may not be representative of a real near side scenario.

When retaining stresses from the *Dynain* file, stresses in some parts of the HBM could not be retained due to a format incompatibility between the *Dynain* dile and *LS DYNA*. This was due to the material type used for certain parts of the model, which were principally in the skin.

4.2 Future work

As future work for this Project, it would be interesting to make further investigation about the retainment and re-initialization of the belt from the pre-crash simulation. Also the whole seat model should be retained, instead of only updated the foam with a reference deformed geometry. Besides, more load-cases (and including intrusions when needed) should be investigated. Comparisons with a full passive model will also be interesting.

In short, after the development of this Project, future work should focus on trying to better replicate the baseline model in order to verify this Project's findings by making more accurate comparisons.

All the models and scripts prepared for this Project (which supposed a great part of it) could be used in order to leave more time for deeper analysis of the results.

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Appendix A

Alignment with Sustainable Development Goals

This Project is focused on the study and of the *SAFER A-HBM v10.0*, which, as a Human Body Model, is a very useful tool when it comes to CAE related with vehicle safety. Therefore, this Project could be principally related with the SDG9 and the SDG10.

SDG9 (Industry, Innovation and Infrastructure) can be related to this Project, as its results may be used to explore different approaches to predict risk injury in order to develop new and/or better safety systems for the coming vehicles, which are moving towards Autonomous Driving. Therefore, it can be considered as an innovation project that could help to the development of the car safety industry.

On the other hand, this Project could also be aligned with SDG10 (Reduced Inequalities). Unfortunately, not every country has the same degree of safety in its vehicles. Achieving the same level of safety systems for vehicles independently of the country where they are sold is one of the main objectives of SUM4all. With the results obtained in this Project, new safety systems will be able to be developed and hopefully they will be implemented in vehicles around the world.

Appendix B

Injury Criteria

An injury criterion is usually a function from which parameters are calculated in order to deduce the severity of injuries in several body parts.

One of the scales in which this injury severity is expressed is the Abbreviated Injury Scale (AIS). This scale, which comes from the Injury Severity Score (ISS), was develpoed by the Association for the Advancement of Automotive Inudstry (AAAM) in 1972.

AIS gives a general rating of a multiple injured patiend, in a 7 level level scale (from 0 to 6), where the higher the number, the higher the severity of the injury [23]. These levels are shown in *Table B.1.*

AIS score	Injury severity
0	No injury
1	Minor
2	Moderate
3	Serious
4	Severe
5	Critial
6	Maximal

Table B.1: AIS scores' injury severity equivalence [23]

Main areas of study in these scale are the head, neck, spine and thorax. A general overview of the injury criteria studied in this Project will follow below.

B.1 Head

The most delicate part of the head, and the one that is usually the most affected, is the brain. The damage to the brain is mainly caused, according to studies, by accelerations applied to the head (regardless of whether they were caused by a direct collision or not), which produce strains in the brain [22].

B.1.1 Head Injury Criteria

Head Injury Criterion (HIC) is one of the most used criteria for estimating skull fracture, which is based in linear accelerations. It is calculated as shown in *Equation B.1*, where $a(t)$ represents the linear velocity of the head, and t_1 and t_2 represent two times separated by a maximum of 15 ms for the HIC_{15}, and by a maximum of 36 ms for the HIC_{36}.

$$HIC = max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,dt\right]^{2.5}(t_2 - t_1) \tag{B.1}$$

A HIC_{36} value higher than 1000 is considered to end up in a fracture skull for a 50^{th} percentile male adult [25].

B.1.2 Brain Injury Criteria

Brain Injury Criteria (BrIC) is another method to predict brain injuries which, contrary to the HIC, is based in angular kinematics. The calculation of BrIC is shown in *Equation B.2*.

$$BrIC = \sqrt{\left(\frac{w_x}{w_{xC}}\right)^2 + \left(\frac{w_y}{w_{yC}}\right)^2 + \left(\frac{w_z}{w_{zC}}\right)^2} \tag{B.2}$$

The parameters w_x, w_y and w_z are the higher rotational velocities registered for the entire time series, while w_{xC}, w_{yC} and w_{zC} are a critical reference value showed in *Table B.2*.

Critical rotational velocity	
w_{xC}	$66.25 rad/s$
w_{yC}	$56.45 rad/s$
w_{zC}	$42.87 rad/s$

Table B.2: Critical rotational reference values [26].

B.2 Neck

Due to the loads applied to the occupant during a crash, the neck is another area that could cause severe injuries. Generally speaking, injuries in the neck tend to be more severe that the ones in other parts of the spine.

B.2.1 Neck Injury criteria

Swift extension-flexion motions of the cervical spine could induce fluid pressure gradients [27], studied by the Neck Injury Criteria (NIC). *Equation B.3* shows how to calculate this injury criterion.

$$NIC(t) = a_{rel}(t) \cdot 0.2 + v_{rel}(t)^2 \tag{B.3}$$

Where $a_{rel}(t)$ and $v_{rel}(t)$ refer to the relative acceleration and velocity, respectively, between the Head CoG[1] and the T1 (first thoracic vertebrae). In order to have an AIS0 score, NIC should never be higher than$15m^2/s^2$ [27].

B.2.2 Nij Neck Injury Criterion

Nij Neck Injury Criterion, developed by the NHTSA, is used for neck injury prediction related with high velocity impacts, as well as for airbag deployment. It is calculated as in *Equation B.4*.

$$Nij = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \tag{B.4}$$

F_z is the axial force (compression/tension) in N and M_y is the bending moment in Nm, both at the neck. F_{int} is the intercept value for compression or tension and M_{int} is the intercept value for extension or flexion at the occipital condyles, both used for normalization [24].

A Nij value of 0.16 predicts a short-term AIS1, while a 0.2 Nij is related with a long-term AIS1 [28].
The rest of the injury analysis was based on predictors already included in the *SAFER A-HBM v10.0* model, which were not standardized injury criterion. They were calculations of maximum strain levels, spine loads and risk of ribs fracture. This is the reason why they were not analyzed in this Appendix.

[1]Center of Gravity